From the Department of Orthopaedics and Traumatology, Helsinki University Central Hospital and University of Helsinki, from the Department of Surgery, Faculty of Veterinary Medicine, University of Helsinki, and from the Institute of Biomaterials, Tampere University of Technology, Finland.

BIOABSORBABLE SELF-REINFORCED POLY-L-LACTIDE SCREWS FOR THE FIXATION OF FEMORAL NECK OSTEOTOMIES IN SHEEP AND FOR CLINICAL PROXIMAL FEMORAL FRACTURES

A clinical and experimental study by
Kirsi Jukkala-Partio

Academic Dissertation

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Supervised by:

Emeritus Professor Pentti Rokkanen, M.D., Ph.D., Ph.D. (Hon. Vet. Med.)

Department of Orthopaedics and Traumatology

Helsinki University Central Hospital

University of Helsinki, Finland

Reviewed by:

Docent Ilkka Arnala, M.D., Ph.D.

Department of Orthopaedics and Traumatology

Kanta-Häme Central Hospital and

University of Kuopio, Finland

and

Docent Peter Lüthje, M.D., Ph.D.

Department of Orthopaedics and Traumatology

Kuusankoski District Hospital and

University of Helsinki, Finland

Opponent:

Docent Olli Korkala, M.D., Ph.D.

Rheumatism Foundation Hospital and

University of Helsinki, Finland

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CONTENTS

ABSTRACT 5

LIST OF ORIGINAL PAPERS 7

ABBREVIATIONS 9

1. INTRODUCTION 11

2. REVIEW OF THE LITERATURE 13
  - 2.1. POLY-L-LACTIDE 13
    - 2.1.1. Chemical properties 13
    - 2.1.2. Biodegradation 14
    - 2.1.3. Biocompatibility 16
    - 2.1.4. Mechanical properties 18
  - 2.2. PREVIOUS STUDIES 19
    - 2.2.1. Experimental studies 19
    - 2.2.2. Clinical studies 21
      - *2.2.2.1. Subcapital femoral fractures 21*
      - *2.2.2.2. Femoral head fractures 22*
      - *2.2.2.3. Clinical studies on bioabsorbable implants 23*

3. THE PRESENT STUDY 27
  - 3.1. AIMS 27
  - 3.2. IMPLANTS 28
    - 3.2.1. Self-reinforced poly-L-lactide screws 28
    - 3.2.2. Other bioabsorbable implants 28
    - 3.2.3. Metallic screws 29
  - 3.3. EXPERIMENTAL STUDIES 29
    - 3.3.1. Material and methods 29
      - *3.3.1.1. Surgical technique 29*
      - *3.3.1.2 Examination methods 31*
  - 3.4. CLINICAL STUDIES 37
    - 3.4.1. General remarks 37
    - 3.4.2. Subcapital femoral neck fractures 37
      - *3.4.2.1. Patients and methods 37*
    - 3.4.3. Femoral head fracture associated with traumatic dislocation of the hip joint 38
      - *3.4.3.1. Patients and methods 38*
    - 3.4.4. Statistical methods 40

4. RESULTS ... 41
4.1. COMPARISON OF THE FIXATION OF SUBCAPITAL FEMORAL NECK OSTEOTOMIES WITH ABSORBABLE SELF-REINFORCED POLY-L-LACTIDE LAG-SCREWS OR METALLIC SCREWS IN SHEEP (I) ... 41
4.1.1.Radiological results ... 41
4.1.2. Strength measurements ... 41

4.2. HEALING OF SUBCAPITAL OSTEOTOMIES FIXED WITH SELF-REINFORCED POLY-L-LACTIDE SCREWS; AN EXPERIMENTAL LONG-TERM STUDY IN SHEEP (II) ... 43
4.2.1. Radiological results ... 43
4.2.2. Histological results ... 45
4.2.3. Microradiographic results ... 48
4.3. BIODEGRADATION AND STRENGTH RETENTION OF POLY-L-LACTIDE SCREWS IN VIVO. AN EXPERIMENTAL LONG-TERM STUDY IN SHEEP (III) ... 48
4.3.1. Radiological results ... 48
4.3.2. Histological results ... 50
4.3.3. Microradiographic results ... 52
4.3.4. Strength retention of SR-PLLA screws in vivo ... 54
4.3.5. Pull-out tests ... 56
4.3.6. Molecular weight measurements ... 56
4.4. TREATMENT OF SUBCAPITAL FEMORAL NECK FRACTURES WITH BIOABSORBABLE OR METALLIC SCREW FIXATION. A PRELIMINARY REPORT (IV) ... 57
4.5. ABSORBABLE FIXATION OF FEMORAL HEAD FRACTURES. A PROSPECTIVE STUDY OF SIX CASES (V). ... 63

5. DISCUSSION ... 67

6. CONCLUSIONS ... 73

ACKNOWLEDGEMENTS ... 75

REFERENCES ... 79

ORIGINAL PAPERS ... 103

ABSTRACT

Bioabsorbable fixation devices have been used during the past two decades in increasing numbers as an alternative to metallic implants in orthopaedics and traumatology. The first bioabsorbable rods and screws in wider clinical use were manufactured from self-reinforced polyglycolide (SR-PGA) and have proved suitable for the fixation of cancellous bone. The development of self-reinforced poly-L-lactide (SR-PLLA) devices with longer strength retention times has also made it possible to use bioabsorbable fixation also on more demanding indications. The degradation time of SR-PLLA implants is long, and although PLLA has been investigated worldwide in experimental as well as in clinical studies, no exact data are available as to the total time of bioabsorption of this implant and of the tissue that it will probably be replaced by when inserted into the bone.

In the present study, the suitability of SR-PLLA screws in the bone fixation in a weight bearing area was examined experimentally and clinically. The degradation and strength retention as well as the bone tissue response were studied in the long-term follow-up radiologically, microradiographically, and histologically and by means of mechanical testing in the experimental part.

SR-PLLA lag screws in the fixation of experimental subcapital femoral osteotomies in sheep were compared with metallic implants. Seven osteotomies were each fixed with two bioabsorbable screws and seven other osteotomies each with two metallic cancellous bone screws. At 12 weeks the sheep were killed and the measurements of the strength was noted. Radiographs at three and 12 weeks showed that union was achieved in six out of seven osteotomies in both groups. At 12 weeks there were no statistically significant differences in load-carrying capacity between the osteotomized bones fixed with SR-PLLA screws and those fixed with metallic screws. These results showed that SR-PLLA screws were strong enough to support the fixation of these weight-bearing bones.

SR- PLLA screws have high initial strength values. At 36 weeks the bending strength of the lag-screws 6.3 mm in diameter which had been implanted in the subcutaneous tissue of sheep had decreased from 257.9 MPa to 36.4 MPa and the shear strength from 131.8 MPa to 19.8 MPa. The pull-out strength of the 6.3 mm lag-screws implanted in the left proximal femur of sheep had decreased from 1507 N to 331 N in 24 weeks.

The biodegradation of SR-PLLA was examined after the implantation of 6.3 mm lag screws in the left proximal femur in nine sheep (follow-up two, three and five years) as well as after the fixation of subcapital femoral osteotomies of ten adult sheep with two 4.5 mm lag screws (follow-up 12 weeks, 1 year, 3 years and 7 years and 4 months). In the plain radiographs all the osteotomies showed bony union. At the bounders between implant and tissue the bone density was already increased in the 12-week control. In the area of the screw channel, the density at

two years was still slightly decreased as compared with the surrounding bone but it gradually increased. On the microradiographs, the increased bone density was seen at one year. At three years, remarkable bone ingrowth was seen in the implant area. At five years, the implant area was not completely replaced by bony tissue but at seven years and four months the bone density had reached the level of cortical bone. Moreover, the computed tomography density values were higher in the area of the screw (mean 2226 HU, SD 205) than in the surrounding cancellous bone (mean 614 HU, SD118). On MRI, the site of the implant was visible as a channel of reduced signal intensity, reflecting sclerotic bone and fibrous tissue. A patchy rim of bone surrounded the channel. The biocompatibility of polylactide proved to be good. There was a mild foreign-body reaction with foam–like macrophages and a few lymphocytes for up to two years. At three years, connective tissue and multinucleated giant cells and macrophages showing a moderate foreign-body reaction surrounded small fragments of the screws. New bone formation and osteoid were also detectable in the screw area. At five years or at seven years and four months, there were no signs of a foreign body reaction and polarized light did not show any birefrigent material. The SR-PLLA screw had been totally biodegraded.

Forty patients with femoral neck fractures were treated by internal fixation using three SR-PLLA lag screws of 6.3 mm in diameter and another 40 patients by using metallic screws 7 mm in diameter, The patients did not differ in age, body weight or primary dislocation of the fracture. In Garden stage I and II fractures there were 5/29 dislocations after SR-PLLA fixation and 8/29 after metallic fixation. In Garden stage III fractures there were 4/9 and in Garden stage IV fractures 2/2 redislocations in both groups. The ability to walk and the range of movement were better after bioabsorbable fixation. SR-PLLA lag screws can safely be used to fix subcapital femoral neck fractures in Garden stages I and II fractures and in younger patients in Garden stage III fractures.

Fracture of the femoral head associated with traumatic dislocation of the hip is a rare but severe injury. Between August 1989 and November 1992 six patients were treated for this at our department by using open reduction and absorbable fixation. In three patients the results were excellent and in one fair. One patient died six weeks after the accident from the consequences of the cerebral injury and one patient was operated on later because of an avascular necrosis of the femoral head. SR-PLLA and SR-PGA implants can be used safely in femoral head fractures and offer a suitable alternative in this intra-articular fracture, because they allow fixation through the cartilage surface without the need of implant removal.

LIST OF ORIGINAL PAPERS

The present study is based on the following papers:

I. Jukkala-Partio K, Laitinen O, Partio EK, Vasenius J, Vainionpää S, Pohjonen T, Törmälä P, Rokkanen P: Comparison of the fixation of subcapital femoral neck osteotomies with absorbable self-reinforced poly-L-lactide lag-screws or metallic screws in sheep. J Orthop Res 15: 124-127, 1997

II. Jukkala-Partio K, Laitinen O, Vasenius J, Partio EK, Toivonen T, Tervahartiala P, Kinnunen J, Rokkanen P: Healing of subcapital femoral osteotomies fixed with self-reinforced poly-L-lactide screws: an experimental long-term study in sheep. Arch Orthop Trauma Surg 122:360-364, 2002

III. Jukkala-Partio K, Pohjonen T, Laitinen O, Partio EK, Vasenius J, Toivonen T, Kinnunen J, Törmälä P, Rokkanen P: Biodegradation and strength retention of poly-L-lactide screws in vivo. An experimental long-term study in sheep. Ann Chir Gyn 90: 219-224, 2001

IV. Jukkala-Partio K, Partio EK, Helevirta P, Pohjonen T, Törmälä P, Rokkanen P: Treatment of subcapital femoral neck fractures with bioabsorbable or metallic screw fixation: A preliminary report. Ann Chir Gyn 89: 45-52, 2000

V. Jukkala-Partio K, Partio EK, Hirvensalo E, Rokkanen P: Absorbable fixation of femoral head fractures. A prospective study of six cases. Ann Chir Gyn 87: 44-48, 1998

The above papers will be referred to in the text by their Roman numerals.

ABBREVIATIONS

AO/ASIF	Arbeitsgemeinschaft für Osteosynthesefragen/Association for the Study of Internal Fixation
ASTM	American Society for Testing and Materials
BS	British Standard
°C	degrees celsius
CT	computed tomography
Dalton	unit of molecular weight (gramme/mole)
dl	deciliter
DSC	differential scanning calorimetry
FA	flip angle
FOV	field of view
g	gramme
g/mol	gramme/mole
GPa	gigaPascal (10^9 N/m^2)
HU	Houndsfield unit
i.m.	intramuscular
K	Kirschner
kg	kilogramme
kGy	kilogray
kV	kilovolt
µm	micrometre
mg	milligramme
ml	millilitre
mm	millimetre
mol	mole
MRI	magnetic resonance imaging
MPa	megaPascal (10^6 N/m^2)
MW	molecular weight
η	intrinsic viscosity (dl/g)
N	Newton (kgm/s^2)
Nm	nanometre
Nm^2	bending stiffness (E*I=bending-modulus multiplied by second moment of cross-sectional area)
PGA	polyglycolic acid or polyglycolide
PLA	polylactic acid (polylactide)
PLLA	poly-levo-lactic acid or poly-L-lactide
ROI	region of interest
s.c.	subcutaneous
SD	standard deviation

SE	spin echo
SR	self-reinforced
T1	longitudinal relaxation time
TE	time to echo
TR	repetition time

1. INTRODUCTION

A new chapter in the history of orthopaedic and traumatological surgery has without doubt been the search for suitable bioabsorbable implants and materials to be used in bone fixation. Decades of development of metallic bone fixation implants have produced devices for almost every indication. Despite their widespread use, metallic implants are not ideal tools for bone fixation because of certain disadvantages. Due to the rigidity of the metallic fixation devices, the physiological load on the bone can be partially transferred to the implant. This so-called stress shielding phenomenon may weaken the bone and lead to temporary osteopenia, disturb the bone remodelling and increase the risk of refractures (Uhthoff and Dubuc 1971, Woo et al. 1976, Tonino et al. 1976, Paavolainen et al. 1978, Slätis et al. 1978, Rosson et al. 1991). Late migration of metallic Kirschner wires and even of screws has been reported (Rai et al. 1991). Corrosion of metallic implants can release metallic ions into the surrounding tissues (Cohen and Wulff 1972, Galante et al.1991, Barbosa 1991). Metallic implants can also interfere with computed tomography (CT) and magnetic resonance imaging (MRI) scans by causing artefacts. Soft-tissue irritation and pain caused by metallic implants are a common reasons for a removal procedure (Jacobsen et al. 1994). To secure the normal growth and to improve complete remodelling of the bone in the pediatric population, metallic implants are routinely removed as soon as bony healing allows (Wolff 1892, Schmalzried et al. 1991). The removal of metallic implants, at least in weight bearing bones, has been recommended as a standard procedure even by the Arbeitsgemeinschaft für Osteosynthesefragen/Association for the Study of Internal Fixation (AO/ASIF) (Müller et al. 1991). There are studies which show that the total cost of the treatment can be reduced by using bioabsorbable implants instead of metallic devices (Böstman 1996, Juutilainen et al. 1997). Also resources like the workload of surgeons and operation theatres that can be saved by avoiding the second operation are available for other demands. Moreover, the psychological advantages of one step fracture care in traumatology or one intervention in orthopaedic surgery without retained hardware or an implant removal procedure later on should be taken into consideration.

To fullfil the criteria of ideal fixation devices, the implants should possess the same rigidity as bone and the rigidity should decrease during the healing process, transferring the load gradually to the bone. During the healing of the bone, the fixation material should decompose and finally disappear totally and the site of the implant should be replaced by new bone. In addition to suitable mechanical properties, there should not be any doubt about the biocompatibility of the material used; the implants should not have any toxic, inflammatory, allergic, carcinogenic or teratogenic side-effects.

To avoid the disadvantages of metallic implants, bioabsorbable polymers have played the main role when alternative fixation devices have been developed since the late sixties (Schmitt and Polistina 1969). The group of alpha-hydroxy acids, such as polyglycolide (PGA), polylactide (PLA) and polydioxanone (PDS) and their copolymers, which have been used as raw material for absorbable sutures worldwide since the seventies (Frazza and Schmith 1971, Ray et al. 1981) have been shown to offer suitable characteristics for bone fixation devices.

The first studies using PLA implants were in the field of maxillofacial surgery (Cutright et al. 1971, Getter et al. 1972). The modest strength values of the early implants led to further development before the wider use, especially in orthopaedic surgery, was possible. The true break-through was achieved in the eighties when Törmälä and his co-workers introduced a new manufacturing method, the so-called self-reinforcing (SR) technique for PGA and PLA implants, in which the bioabsorbable polymeric matrix was reinforced with oriented, fibrous reinforcing elements, which had the same chemical composition as the matrix (Törmälä et al. 1987, Törmälä 1992). This led to a considerable improvement in the mechanical properties, and in 1984 the first internal fixation of a malleolar fracture by using SR-PLLA/PGA copolymer rods was carried out (Rokkanen et al. 1985). SR-PGA screws came into clinical use in May 1987 and SR-PLLA screws in July 1988. Since then, new implant designs, such as tacks, wires, plugs, plates and arrows have been developed (Rokkanen et al. 2000).

To be able to give the necessary support to the bone during healing, the fixation device should have sufficient initial strength and stiffness and an appropriate retention rate of these values. The fixation device can be exposed, not only to simple shear load, but also to tensile, bending, pullout, torsion and compression forces. Small cancellous bone fractures with shorter healing times can be treated with good results by using implants with lower initial strength values and faster degradation, but the level of requirements is especially high when concerning fractures or osteotomies which are exposed to great mechanical stresses or in which the time of consolidation is known to be long.

Many biocompatible, bioabsorbable polymers are now available for both experimental and clinical use and, as a result of intense research, there have been numerous studies investigating their biomechanical properties. However, in addition to the lack of detailed descriptions of different materials and the manufactory of the samples and variety of testing conditions (in vivo/ in vitro), the strength measuring methods have varied widely in different studies making the reports not exactly comparable (Daniels et al. 1990, Claes 1992). To give adequate information about bioabsorbable implants, not only the characterization of the material and initial strength values, but also the rate of degradation of mechanical properties in tissues, is essential. If standardized methods have not been used, the test conditions should be documented in detail.

The purpose of the present study was to analyze two aspects; to find out whether the mechanical properties of bioabsorbable self-reinforced poly-L-lactide implants are suitable for the fixation of fractures and osteotomies that are exposed to great mechanical stresses, and to elucidate the degradation and in vivo strength retention of SR-PLLA lag-screws as well as the tissue response to the implant material in the long term.

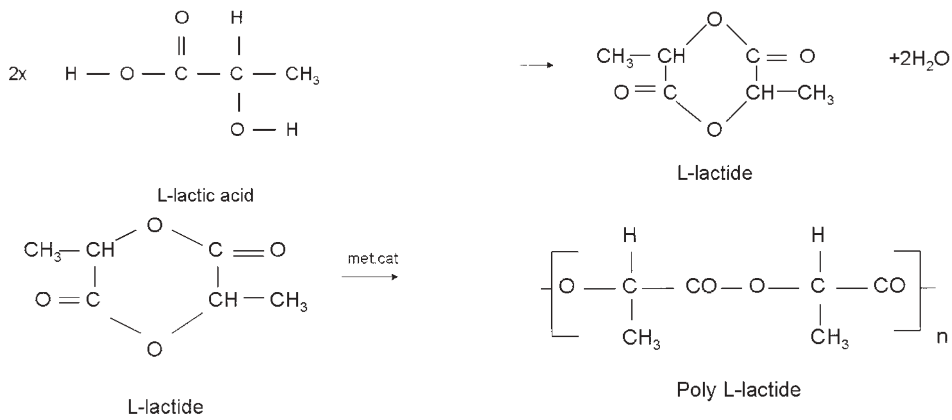
2. REVIEW OF THE LITERATURE

Several polymers have been found to be bioabsorbable in living tissue. Aliphatic polyesters (polymers and copolymers) of alpha-hydroxy acid derivatives are today the most important materials in bioabsorbable osteosynthesis devices, whereas polyglycolide (PGA) and polylactide (PLA) and their copolymers offer the best properties for orthopaedic devices, as being the strongest polymers of this class (Vert et al. 1981, Törmälä 1992). According to Higgins (1954) Bischoff and Walden synthesized polyglycolide with a low molecular weight in 1893. PGA with a high molecular weight and plastic properties was reported by Higgins in 1954, whereas a high molecular weight polylactide was introduced one year later by Schneider (1955). In 1966 Kulkarni et al. showed in animal experiments that bioabsorbable poly-L-lactide (PLLA) can be used as the synthetic raw material of surgical implants and since then many investigations have been made in this field by different research groups (Kulkarni et al.1971, Schmitt and Polistina 1969, Cutright et al. 1971,1972,1974, Vert et al. 1981).

2.1. POLY-L-LACTIDE

2.1.1. Chemical properties

Lactide is the common name for the cyclic diester form of monomeric lactic acid that exists in two enantiomeric forms: optically active stereoisomers L(+)-lactic acid and D(-)-lactic acid, with similar intrinsic chemical properties but opposite configurational structures. Therefore three different lactides can be formed: L(-)-lactide, D(+)-lactide and the optically inactive meso-lactide. The racemic 1:1 mixture of L- and D- is generally called D,L-lactide. For the production of polymers, L- and D,L-lactide are used almost exclusively. Polymers with high molecular weights suitable for orthopaedic devices are exclusively produced by the anionic ring opening polymerisation of the cyclic lactides under the influence of a low catalyst (inorganic metal salt) concentration (Cutright et al. 1974, Vert et al. 1981, 1984, Lowe 1954, Schneider 1955, Jamshidi 1984, Bendix 1998). Poly-(L-lactide) (PLLA) is a stereoregular homopolymer derived from pure L-lactide (Fig. 1). PLLA has until now been more popular for clinical use in orthopaedics because of its high initial strength and long strength retention time, but intensive investigations have been made to develop different stereocopolymers of L- and D- isomers, where physical and chemical properties, such as strength retention or degradation time can be brought to more optimal level for indications with different demands, which occurs by changing the ratio of L- and D-lactic acid repeating units in the polymer chain. Poly(D,L-lactide) (PDLLA) obtained from racemic D,L-lactide mixture is an intrinsically amorphous, glassy polymer with relatively low mechanical properties and has accordingly little use in orthopaedics. Copolymers of L- and D,L-lactide are amorphous when the percentage L-stereoisomer is lower than 90 mol%. By adding small amounts of the D-stereoisomer to the L-chain, e.g. 5% in a 90:10 copolymer, the crystallinity is significantly lowered, compared to the pure PLLA. This combines improved biocompatibility of the material with good mechanical properties (Bendix 1998). PLLA is a pale, semicrystalline, and thermoplastic polymer. A highly crystalline PLLA with the molecular weight of over 100 000 has its melting point at 174 to 184 °C with a glass transition temperature of 57 to 58 °C (Vert et al. 1981, Jamshidi 1984, Törmälä et al. 1998).



met. cat. = metal catalyst e.g. Zn powder or tin octoate

Figure 1. Synthesis and ring opening polymerisation of L-lactide

2.1.2. Biodegradation

Biodegradable aliphatic polyesters undergo a two-stage degradation process in the living tissue. In the first lengthy phase, water splits the chemical bonds of poly-L-lactide chains to lactide mainly by a non-specific hydrolytic de-esterification. In the second phase, lactide becomes incorporated into the mitochondrial citric acid cycle and is subsequently excreted by the lungs as carbon dioxide and through the urine as water (Miller et al. 1977, Williams 1979, Hollinger 1983, Hollinger and Battistone 1986) (Fig. 2).

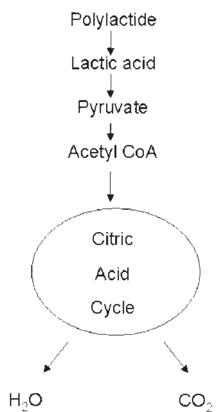


Figure 2. Biodegradation of polylactide (Kulkarni et al. 1966, Lehninger 1982, Williams 1982, Hollinger and Battistone 1986).

The degradation can also be accelerated by enzymatic activation to a lesser extent. Especially pronase, proteinase-K and bormelain are mentioned as taking part in the process, but also fincin, esterase, and trypsin, whereas the presence of lactate dehydrogenase shows negative results (Williams 1981, Hollinger and Battistone 1986, Schakenraad et al, 1990).

At the cellular level, polylactide degrades first to small particles. The particles trigger the phagocytosis reaction, by which macrophages and multinuclear giant cells digest the polymer fragments 10-80 μm in diameter (Woodward et al. 1985, Bos et al. 1991, Rozema et al. 1994).

The degradation time of bioabsorbable polymers in living tissue depends on microstructural (chemical composition of material, molecular weight, crystallinity, molecular orientation, impurities, matrix and reinforcement morphology, porosity and surface quality), macrostructural (size and geometry of the implant, weight/surface ratio) and environmental factors (site of implantation in living tissue) (Leenslag et al. 1987, Vert et al. 1992, Törmälä et al.1998). Therefore, a comparison of the results of studies concerning strength retention and biodegradation can lead to misinterpretations because the polymeric raw materials and the manufacturing procedures are not well characterized and the physiochemical parameters of the samples differ. The high initial molecular weight and high degree of purity make the degradation rate of polylactide slower (Vert et al. 1981, Nakamura et al. 1989, Pistner et al. 1993, Zhang et al. 1994). Also a highly crystalline polymer can resist the hydrolysis better than an amorphous polymer in which the monomers are more loosely packed (Gilding 1981, Vert et al. 1984). Sterilisation methods contribute to MW and mechanical properties of PLLA whereas high-energy irradiation, the low temperature-cycle ethylene oxide process, and steam and standard heat treatments tend to reduce the strength values more than low temperature sterilisation under vacuum or a dry inert gas (Gogolewsky and Mainil-Varlet 1996). The biodegradation of the bioabsorbable implants is usually faster in bone than in soft tissue (Vasenius et al. 1990).

The ultimate degradation time of PLLA implants is several years. There are certain limitations in long-term studies, because the life-time expectancy of most laboratory animals is limited and the potential examination methods for clinical studies are few, especially when the implants are inserted into the bone. In an experimental study on rats Matsusue et al. (1995) found that PLLA rods manufactured by using a drawing technique were absorbed almost completely in 62 months when implanted intramedullarily and completely in 69 months after subcutaneous implantation. In another experimental study SR-PLLA screws were inserted axially into the distal femurs in 18 rabbits. After three and 4.5 years follow-up, birefringent polymeric material was still present in the center of the implant channel. Only sparse reactive cellular activity was seen at the tissue-implant boundary. In transmission electroscopy specimens at three years, spherical and polygonal particles with an average area of two µm≤ were seen to be located intracellularly within phagocytic cells, filling them to a great extent. In the 4.5 year's specimens, the size of the polymeric particles was significantly smaller ($p<0.02$) (Laitinen et al. 2002). In long-term clinical studies, remnants of PLLA could still be detected after 5.7 years of implantation (Bergsma et al. 1995), at five and a half years (Tams et al. 1996) and after five years (Suuronen et al. 1998) when placed on the bone.

2.1.3. Biocompatibility

The biocompatibility of PLLA materials has been investigated in many experimental and clinical studies and PLA is generally well tolerated by the host tissue (Majola et al. 1991, Matsusue et al. 1997, Nordström et al. 1998). However, owing to the unexpected foreign-body reactions in the early clinical studies and in clinical use dealing with PGA implants, special care is taken when introducing new materials into clinical use. Cutright and Hunsuck (1972) operated twelve experimental orbita fractures in Macaca rhesus monkeys by inserting PLA sheets of 1.5 mm in the orbital floor. In histological examinations at 4,8,12,18,24 and 38 weeks, the resorption of PLA was found to be accomplished by a peculiar phagocytic process involving phagocytic cells, giant cells, and villous projections, but the biocompatibility was very good and no lymphocytic or plasmacytic infiltration was present after 8 weeks. Eitenmüller et al. (1996) reported on 19 ankle fractures fixed by PLLA plates and screws. Fracture union was achieved in six weeks, but 52% of the patients developed an aseptic soft tissue problem one year after implantation. The geometry of the plates and screws were changed by reducing the volume and flattening the screw heads. In the second protocol with seven patients, no soft tissue reactions were shown.

The first 1043 clinical operations in orthopaedic and trauma patients, using SR-PLLA implants, were analyzed in our department (Juutilainen et al. 2002). There were 21 infections (2%) but no sinus formation. Fluid accumulations (two of them small) were observed in three patients, indicating that the implants were not fully introduced into the bone.

Biocompatibility and degradation behaviour of SR-PLLA screws in 51 displaced ankle fractures were studied clinically and radiographically, using CT scans. The follow up time was at least three years (mean 52 months). In addition, biopsy specimens for histological examination were taken in five patients. An accurate position, with union, was maintained in 50 patients. A mild transient subcutaneous late foreign body reaction occurred in one patient 22 months after fixation. In three patients a disturbing palpable uncut head of the screw was removed. The incidence of foreign body reactions to SR-PLLA screws in the ankle fractures seemed to be low, but the degradation process considerably slower than anticipated (Böstman et al. 1995).

In a long-term study of dislocated ankle fractures fixed with SR-PLLA implants, in five out of 16 patients, a late tissue reaction with mild symptoms was observed over the uncut extruding screw heads in medial malleoli. The swelling appeared late (40-115 months after primary surgery). There were no signs of bacterial infection, but the prominent masses caused discomfort and all five patients agreed to undergo operation to remove the palpable masses. The histological and immunohistochemical examinations showed mild to moderate foreign body reactions, but inflammatory cells were almost absent and the reaction was surrounded by a capsule of connective tissue (Voutilainen et al. 2002).

Tegnander et al. (1994) reported of diffuse swelling in 6 out of 10 knees after arthroscopic fixation of osteochondritis dissecans with 2 to 5 SR-PLLA pins. The reaction occurred in one patient as early as 3 days after the operation. The author was not aware of the size of the implants and he was not familiar with the operation technique – especially if the pins were sunk under the cartilage surface (personal communication). Recently, when the proper operation techniques have been followed, this kind of complications have been rare and caused by mechanical soft tissue irritation by a bulky screw head or by breakage or loosening of the implant in the joint (Takizawa et al.1998).

In a study of first 2500 SR-PGA (n=1879) and SR-PLLA (n=621) fixations with emphasis on the knee there were no clinical tissue responses in SR-PLLA fixations and the incidence when using SR-PGA implants in the knee was 4.3% (n= 92) with four fluid accumulations and one synovitis, which was treated by aspiration with a good final outcome (Tuompo et al. 2001).

In an experimental study on rats, SR-PLLA and SR-PGA pins 1.5 mm in diameter were implanted intra-articularly in the distal femur between the intercondylar notch 1 mm above, on the same level or 1 mm under the articular cartilage surface, as well as directly in the bone extra-articularly. In general, the tissue responses were mild in the follow-up at 3, 6, 12 and 24 weeks. In the histologic analysis the most favourable implantation depth was found to be under the surface, especially in intra-articular implantation. In these specimens the orifice was covered with new trabecular bone at three weeks (Koskikare et al. 1998).

In a comparative study of infections 3111 ankle fractures were treated with metallic (2073 patients) and bioabsorbable SR-PGA or SR-PLLA implants (1012 patients). The infection rate associated with metallic fixation was 4.1%, compared with 3.2% absorbable fixation (p 0.3). Deep infections were equally common with both fixation methods (0.4%) (Sinisaari et al. 1996).

Oppenheimer et al. (1955) showed foreign-body tumour genesis after long-term implantation of any polymer in rodents. This phenomenon, nowadays known as the Oppenheimer effect, was observed already by Turner (1941) after subcutaneous implantation of disks of bakelite in rats and studied further especially by Oppenheimer et al. (1955), Ott (1970) and Brand et al. (1975). In rats and mice, an incidence of up to 80% sarcomas could be achieved by subcutaneous implantation of chemically inert materials and depends not on the chemical structure of the implant but on its form, size and time. The incidence in rodents depends on the amount of avascular cicatricial tissue (Ott 1970). Chronic inflammation inhibits this kind of tumour formation (Brand et al.1975). Nakamura et al. (1994) implanted 50 PLLA plates (20x10x1mm) subcutaneously into 50 rats and 50 plates of medical-grade polyethylene of the same size into the control group. Additionally 30 rats were sham operated. Mesenchymal tumours arose in 23 out of 50 rats with PLLA plates and in 22 out of 50 control rats within a two-year observation period. In 30 rats given sham operations, no tumours appeared.

Pistner et al. (1993) studied the degradation of three poly-L-lactides with different molecular weights. Small blocks (3x3x2mm) and rods (25x3x2mm) were implanted into the muscle and subcutaneous tissue of 70 rats for one to 116 weeks. Three rats developed a foreign body sarcoma in direct contact with bigger rods independent of the different molecular weights. In one experimental in vitro study, normal fibroblasts and epithelial tumour cells were challenged by exposure for 24 hours to extracts of a pure metal, a metal alloy, and PLLA to assess cellular growth effects. While none of these materials significantly altered fibroblast growth rates, PLLA inhibited carcinoma cell growth at 5.0% extract concentration. No carcinoma cell growth effects were seen from the exposure on stainless steel or commercially pure titanium extracts (Campbell et al.1994). Tumour genesis around foreign body material has been described as a tendency of rodents and has not been detected in human patients.

2.1.4. Mechanical properties

The first polylactide implants were manufactured by traditional melt moulding techniques (such as extrusion and injection moulding). However, the absorbable polymers as such are either brittle or too weak and flexible for safe clinical use. The bending strength values of non-reinforced absorbable polymers are typically between 40 and 140 MPa (Törmälä et al. 1998). These are below the strength of cortical bone, for which bending strength values of 180-195 MPa, shear strength of 68 MPa, and bending modulus (also known as Young´s modulus or elastic modulus) 9.5-11 GPa have been measured (Reilly and Burstein, 1975 Tonino et al. 1976).

It was found that bioabsorbable polymer matrices could be reinforced with bioabsorbable fibres to increase their strength, ductility and bending modulus. Christel et al. (1980) and Vert et al. (1981) developed PLLA plates reinforced with PGA fibres. The reinfocement increased the bending modulus from four to six GPa. Other reinforcing material combinations were also used by many research groups. Lin (1986) introduced analogous unidirectional composites of PLLA matrix reinforced with calcium/phosphorous oxide (CaP) based glass fibres showing a bending strength of 161.3 MPa and a bending modulus of 27.0 GPa but a shear strength of only 19.2 MPa, which might be due to the brittle nature of CaP.

The research group of Törmälä and Rokkanen selected the so-called self-reinforcing technique to develop ultra-high strength, bioabsorbable implant materials (Pohjonen et al.1989). Self-reinforcing means that the bioabsorbable polymeric matrix is reinforced with oriented, fibrous reinforcement elements, which have the same chemical composition as the matrix and are bound together under elevated pressure and at a consistent temperature. The original sintering technique (Törmälä et al. 1988) has been changed to so-called partial fibrillation by orientational drawing (Törmälä et al. 1990). The maximum bending strength and modulus obtained for the SR-PGA rods are 415 Mpa and 18 GPa (Törmälä 1992) and for SR-PLLA screw 300 Mpa and 10 GPa, respectively (Törmälä 1993) (Table 1).

Table 1. Mechanical properties of poly-L-lactides and cortical bone according to the literature.

	Bending strength (MPa)	Shear strength (MPa)	Bending modulus (Gpa)
PLLA self-reinforced	200-300, ref. 1, 2, 3, 4	94-220, ref. 1, 2, 4	7-10, ref. 1, 2, 4
PLLA non-reinforced	113-145, ref. 1, 3, 5, 6, 7	53-68, ref. 1, 3, 6, 7	3-5, ref. 1, 3, 5, 7
Cortical bone	180-195, ref. 8	68, ref. 9	9,5-11, ref. 8

References: 1= Törmälä 1992,2= Majola et al. 1992, 3=Pohjonen and Törmälä 1996, 4= Törmälä et al. 1998, 5 =Gerlach and Eitenmüller 1987, 6= Pohjonen et al. 1988, 7=Mainil-Varlet et al.1997, 8=Tonino et al.1976, 9= Reilly and Burstein 1975

2.2. PREVIOUS STUDIES

2.2.1. Experimental studies

In the field of weight-bearing bones, the fixation properties of several bioabsorbable devices has been tested. In one experimental study on sheep, SR-PLLA screws were compared with metallic screws. Nine osteotomies of the neck of the mandibular condylar process were fixed with SR-PLLA screws and nine with metallic screws .Two of the osteotomies were left unfixed. The consolidation of the fixed osteotomies was uneventful in both groups but seemed to be faster in the polylactide group. Malalignment was detected at three and 12 weeks control in both unfixed condyles (Suuronen 1991). In another study, 36 distal osteotomies of rabbit femur were fixed by using SR-PLLA screws and 36 osteotomies with SR-PDLLA/PLLA screws. In the SR-PLLA group 34/36 and in the SR-PDLLA/PLLA group 31/36 osteotomies healed without delay or angular deformity (Majola 1991). Manninen and Pohjonen (1993) used SR-PLLA rods 4.5 mm in diameter in the intramedullary fixation of 42 tibial cortical bone osteotomies in rabbits. None of the rods broke during the follow-up period of three to 48 weeks. There were two non-unions. Radiological, histological, microradiographical and oxytetracyclic fluorescence studies showed normal healing in the specimens where union had occurred.

A large number of investigations have been performed on the effects of electromagnetic fields on bone formation. Bassett (1968) was the first who found out that piezoelectricity has a certain influence in the remodeling and resorption of the bone. In 1996 Ikada et al. showed in an experimental study that the piezoelectric constants of PLLA films increased with the draw ratio and after passing a maximum at a draw ratio around 5, decreased. The PLLA rods were implanted intramedullarily in the cut tibiae of cats for internal fixation up to 8 weeks. Fracture healing was clearly promoted with increased callus formation as the draw ratio of the PLLA rod increased, whereas the undrawn PLLA as well as a polyethylene control rod had no effect on callus formation, or rather, retarded it. This finding suggests that the promotion of fracture healing by fixation with drawn PLLA can be ascribed to the piezoelectric current generated by the strains accompanying leg movements.

Pullout strengths of three different non-reinforced PLA screws machined from a block and with an outer diameter of 2.6-3.4 mm and three different metallic screws with outer diameters of 2.0 to 2.7 mm were tested in vitro in porcine ribs. The pullout force of the PLA screws was between 178 and 326 N and that of the metallic screws varied from 318 to 431 N (Wittenberg et al. 1991). Pullout strengths of SR-PLLA rods were compared with those of metallic Kirschner wires in an experimental study (Rubel et al. 2001). Ten absorbable rods and 10 K-wires, both 2 mm in diameter were inserted into young bovine cancellous bone and then pulled out using a material testing machine. The K-wire mean pullout force was 37.7 N (SD 13.6) and the SR-PLLA rod pullout force was 53.6 N (SD 19.3). Significant differences ($p<0.01$) favoring bioabsorbable pullout strength were noted.

In another study, the biomechanical properties of stainless steel and bioabsorbable fixation of the clavicle to the base of the coracoid were evaluated. Seven matched pairs of fresh frozen cadaver shoulders were prepared by removing all the soft tissue except the acromioclavicular and coracoclavicular ligament complexes. The shoulders were fixed with 4.5 mm stainless steel screws, while the contralateral shoulders were fixed with 4.5 mm PLLA screws. The average pull-out strength of the metallic screws was 720.6 +/-244.9 N, which was not statistically different from that of the bioabsorbable screws of 580.4 +/- 188.6 N. Both of these fixations exceeded the reported strength (500N) of the intact coracoclavicular complex (Talbert et al. 2003).

In one study, a transverse subcapital osteotomy was created with a hand saw in 19 pairs of human cadaver femora. Two fixation methods were randomly used in each pair: three SR-PLLA screws or three standard metallic cannulated screws with outer diameter of 6.3 mm and 6.9 mm, respectively. Fixations were exposed to a progressive cyclic loading test to determine the deflection curves. The average maximum load-carrying capacity was 3400 N for metallic and 2600 N for absorbable fixation. The SR-PLLA screws showed sufficient strength for considering clinical trials (Vasenius et al. 1998).

2.2.2. Clinical studies

2.2.2.1. Subcapital femoral fractures

The rapid growth of the elderly population has led to a related increase in the incidence of osteoporotic fractures such as those in the hip region. In 1990, an estimated 1.3 million hip fractures occurred and approximately half of these fractures were femoral neck fractures (Gullberg et al. 1997). Fractures of the hip were earlier considered as untreatable, and until last century little was done for them. In 1878, von Langenbeck performed the first open reduction and internal fixation for a fractured hip and in 1897, Nicolaysen reported 13 cases in which a steel spike was used. The treatment of cervical femoral fractures gained more popularity when, in 1931, Smith-Petersen published data of the triflanged nail, which he had developed especially for this indication. The first idea of multiple pinning was introduced by Moore in 1937. Shortly thereafter, metal substitutes were devised for the disunited head of the femur.

At present these fractures are almost always treated operatively, though, non-operative treatment of impacted femoral fractures (Garden I) has also been advocated (Raaymakers and Marti 1991). In more recent reports, the rate of nonunion of conservatively treated undisplaced intracapsular fractures has been 40% or more (Bédat et al. 1997). Comparative studies of conservative and operative treatment of undisplaced fractures suggested that operatively treated patients had a lower risk of complications in healing (Cserhati et al. 1996). Several major investigations have been made in Nordic countries during the last few decades (Lüthje 1983, Nilsson 1989, Husby 1990, Kuokkanen 1992). In these studies the importance of operative treatment of femoral neck fractures has been documented. The standard treatment for femoral neck fractures is either prosthetic replacement or internal fixation, whereby in Nordic countries internal fixation is conventionally used, especially in younger patients and in older patients when the fracture is not displaced. The goal has been to preserves the patient´s own femoral head to retain the physiological function of the hip joint and because of the risk of the need for a later revision procedure. In elderly patients the arguments against arthroplasty diminish with a shorter life expectancy.

The rate of displacement of the fracture has a significant influence to the outcome of the treatment. Roden et al. (2003) compared in a randomised study 100 displaced femoral neck fractures (Garden III-IV) treated either with screws or with bipolar hemiprostheses. After a minimum follow-up of five years, 34 out of 53 patients were reoperated on in the internal fixation group and three out of 47 patients in the prosthesis group. Dislocation of the prosthesis occurred in 7 out of 47 patients, all within four months. In another prospective randomised multicentre trial, 409 patients, aged 70 years and over, with subcapital femoral neck fractures Garden stage III or

IV were treated with internal fixation or arthroplasty. After two years, the rate of failure was 43% in the internal fixation group and 6% in the arthroplasty group ($p<0.001$) (Rogmark et al. 2002).

Numerous classifications have been created to discover a suitable treatment (Pauwels 1935, Garden 1961), but they have all proved unsatisfactory in terms of predisposing the outcome (Nieminen and Satokari 1975, Frandsen et al. 1988). Blundell et al. (1998) investigated the intra- and interobserver accuracy and value in predicting treatment of the AO classification. It was found to show very poor intra- and interobserver reliability and thus its value in predicting the outcome of treatment has remained limited. The authors concluded that a simplified system in which the subdivisions were allocated to one of the three groups of undisplaced, displaced, and basal fractures should be of more value. In addition, other useful predictive criteria (unfavourable outcome: small head fragment, comminution of the femoral calcar, and varus angulation of the head) have been studied to help in choosing the best treatment (Alho et al. 1992).

For internal fixation, different devices have been used during the last few decades. The fixation method has been mainly the sliding-hip-screw or three-hip-screw technique, which has provided a strong fixation (Rehnberg and Olerud 1989, Nilsson 1989, Husby 1990, Kuokkanen 1992). In the Garden I-II fractures the results have been comparable between the three-screw fixation and the screw-angle plate techniques (Kuokkanen 1992). In a meta-analysis (Parker and Blundell 1998), all the randomized trials were reviewed, comparing different implants for treating intracapsular fractures of the hip. Twenty-five randomized trials were identified involving 4925 patients. Screws appeared to be superior to pins. It was not possible to determine the optimum number or type of screws. An implant with side-plate showed no advantages.

To our knowledge, bioabsorbable fixation has not been performed earlier in the treatment of subcapital femoral fractures.

2.2.2.2. Femoral head fractures

A traumatic dislocation of a hip joint with the fracture of a femoral head is a rare but severe injury. This Epstein type V fracture-dislocation (Thompson and Epstein 1951) has been reported in the literature to have an incidence ranging from 6 to 16 percent of all hip dislocations (Epstein 1973, Roeder and DeLee 1980, Butler 1981, Lang-Stevenson and Getty 1987, Hougaard and Thomsen 1988). Femoral head fractures associated with dislocation of the hip joint are divided into four types according to the classification of Pipkin (1957) (Table 2).

Table 2. Classification of femoral head fractures according to Pipkin (1957).

Type 1	Type 2	Type 3	Type 4
Dislocation with fracture of the femoral head caudad to the fovea capitis femoris	Dislocation with fracture of the femoral head cephalad to the fovea capitis femoris	Type 1 or 2 associated with a fracture of the femoral neck	Type 1 or 2 associated with a fracture of the acetabular rim

Because of the risk of avascular femoral head necrosis, every traumatic dislocation of the hip must be treated as an emergency procedure. Immediate reduction gives a better prognosis than when done later (Epstein et al., 1985 and Hougaard and Thomsen, 1988). The methods of recommended treatment have varied from primary closed or open reduction without fixation (Pipkin 1957, Kelly and Yarbrough 1971) to excision of fragments (Epstein 1974, Epstein et al. 1985, Jakob et al.1987, Keene and Villar 1994) or even primary hip replacement or arthrodesis especially in type III fractures, although the internal fixation has been the most recommended treatment (Roeder and DeLee 1980, Mowery and Gershuni 1986, Murray et al. 1988, Warren 1991, Swiontkowski et al. 1992).

Because of the rarity of this specific injury, the reported series are small and there are no prospective randomized studies. It has been shown retrospectively that primary open reduction (Epstein 1973, Epstein et al. 1985) especially with fixation (Lang-Stevenson and Getty 1987), should be preferred to closed treatment. Some authors recommend immobilization or even traction of the extremity after the treatment procedure with or without fixation.

The technical difficulty in the fixation of these fractures by using metallic implants is that the fragments usually consist of a relatively large part of the cartilage surface with a small bony part of the femoral head, which is very difficult to handle with a retrograde fixation. Most of the metallic fixations in the literature have been performed that way extra-articularly, but even intra-articular fixation with metallic screws by sinking the screw head under the cartilage surface has been reported. Removal of metallic implants after fracture healing has its risks. Therefore, osteosynthesis material has usually been left in place, in case it is not necessary to remove it because of migration. Huang et al. reported in 1995 of femoral head fractures operated on from 1993 to 1995, using SR-PLLA devices. The results were promising.

2.2.2.3. Clinical studies on bioabsorbable implants

Since the self-reinforced PGA/PLA rods were introduced into clinical use in November 1984 (Rokkanen et al. 1985), many clinical series of the bioabsorbable fixation of ankle fractures have been published (Böstman et al. 1987, Ruf et al 1990, Partio et al.1992a, Pihlajamäki et al. 1994). SR-PLLA screws have been available for clinical use since August 1988 when the first ankle fracture was operated on using the new implant (Partio 1992). In one study 155 patients with a closed displaced ankle fracture were treated with medial malleolar fixation with the use of either 4.0 mm orientruded polylactide screws (83 patients) or 4.0 mm stainless steel screws (72 patients) combined with metallic fixation of the lateral malleolus. It was concluded that polylactide is a safe and effective alternative to stainless steel on that indication (Bucholz et al. 1994). Bioabsorbable screws, pins and nails made of orientated ultra high strength PLLA were used for fixation of bone grafts (84 patients), fractures (49 patients) osteotomies (8 patients) and in 2 other indications. Bony union was achieved in all except one case (Yamamuro et al. 1994).

In the lower extremity, the most common operations with bioabsorbable fixation beside ankle fracture treatment have been Chevron osteotomy, proximal first metatarsal osteotomy and arthrodeses of the first metatarsophalangeal joint (Hirvensalo et al. 1991, Niskanen et al. 1993, Burns 1995, Barca and Busa 1997, Voutilainen et al. 2002). Displaced intra-articular talar and calcanear fractures have been treated with SR-PLLA and SR-PGA screws and rods, with equal results compared to previous operative methods utilizing various metallic implants (Kankare and Rokkanen 1998, Kankare 1998). Other more mechanically demanding fixations have also been performed, such as talocrural and subtalar arthrodeses (Partio et al. 1992b, 1992c, Juutilainen and Pätiälä 1995), proximal tibial fractures and osteotomies (Kankare 1997, Tuompo et al. 1999) and distal epiphyseal femur fractures (Partio et al. 1997). The use of PLLA screws in rotational acetabular osteotomy in 41 dysplastic hips of patients with an average age of 32 years (range 12-55 years) has been reported. After a minimum follow up of 6 months, bony union occurred uneventfully in all cases (Nakamura et al. 1999).

In the upper extremity, bioabsorbable fixation has proved to be sufficient in fractures of the distal humeral epiphyses (Böstman et al. 1989, Mäkelä et al. 1992), humeral epicondyle (Partio et al. 1996), olecranon fractures (Hope et al. 1991, Partio et al. 1992d, Juutilainen et al. 1995), distal radial fractures (Hoffman et al. 1989, 1992, Casteleyn et al. 1992) and fractures of the hand (Kumta et al. 1992, Pelto-Vasenius et al. 1996).

Bioabsorbable screws have also been combined with metallic implants in severe ankle fractures with a ruptured tibio-fibular syndesmosis to eliminate separate removal of the transfixing material. In one study, seven patients were treated by an ordinary metallic plating and SR-PGA screw transfixation of syndesmosis. All the patients ended up with an excellent or good result (Korkala et al. 1999). In another study of ankle fractures with ruptured syndesmosis, 18 patients with a bioabsorbable SR-PLLA transfixation screw combined with metallic devices and 12 patients with metallic fixation were examined after a minimum follow-up period of 12 months. There were no significant differences between the patient groups in radiographic and CT films, loaded dorsal range of movement of the ankle, or duration of sick leave (Sinisaari et al. 2002).

Bioabsorbable implants have been used successfully in the intra-articular indications, like fractures of the radial head (Becker 1988, Hirvensalo et al. 1990) or in the fixation of osteochondritis dissecans in the knee joint (Matsusue et al 1996, Tuompo et al. 2000) where the possibility to the fixation through the articular surface has also been shown to have its benefits. In the first studies, PGA was used as fixation material, but later PLLA has gained more popularity, especially in intra-articular indications, because of its degradation behaviour with a mild or missing foreign-body reaction described for PGA. In the last few years the increasing interest in arthroscopic surgery has made the use of absorbable screws and anchors almost the method of choice in knee and shoulder surgery (Burkhart 2000, Boileau et al. 2002, Barber et al. 2003).

3. THE PRESENT STUDY

3.1. AIMS

The aims of the study were to find answers to the following questions:

1. Are the mechanical properties of SR-PLLA lag-screws sufficient for the fixation of subcapital femoral osteotomies in sheep and how does a subcapital femoral osteotomy fixed with SR-PLLA lag-screws heal in sheep? (I, II)

2. How fast is the biodegradation of SR-PLLA lag-screws in the upper femur of sheep and what is the tissue response to the implants?? (II, III)

3. How fast is the strength retention of SR-PLLA lag screws in bony tissue and in subcutaneous tissue of sheep? (III)

4. Is it possible to treat subcapital femoral neck fractures safely with bioabsorbable SR-PLLA lag screws? (IV)

5. Can bioabsorbable fixation devices be used in the treatment of femoral head fractures? (V)

3.2. IMPLANTS

3.2.1. Self-reinforced poly-L-lactide screws

The bioabsorbable lag-screws used in this study were manufactured of self-reinforced (SR) poly-L-lactide by Bioscience Ltd. (at present Linvatec Biomaterials Ltd.), Tampere, Finland. The raw material was purified medical grade PLLA with a viscosity average molecular weight (MW) of 710 000 daltons (Purac Biochem BV, Gorinchem, The Netherlands). After melt extrusion and solid state self-reinforcing using the die drawing method, the MW was 220 000+/-20 000 Daltons. A minimum dose of 25 kGy (Kolmi-Set, Ilomantsi, Finland) gamma irradiation was used to sterilize the screws, and that further decreased the MW of SR-PLLA implants to 50 000+/-5 000 Daltons. The percentage of crystallinity of the implants was 72 +/- 5 % as determined by differential scanning calorimetry (DSC) measurements. The SR-PLLA implants had the following mechanical properties: bending strength 175-225 MPa, bending modulus 6.5-7.5 GPa, bending stiffness 0.23-0.26 Nm^2, and shear strength 115-135 MPa. The nominal outer diameter of the smaller lag screws were 4.5 mm and the inner diameter 3.5 mm (I,II,IV). In the bigger lag- screws the values were 6.3 mm and 4.5 mm respectively (III and IV). The length of the screws in the experimental studies was 40 mm (inserted in the proximal femur) and 70 mm (subcutaneous implantation) and in the clinical study of femoral neck fractures (IV) they varied from 75 mm to 110 mm. The length of the threaded part was 20 mm (Fig. 3).

Figure 3. SR-PLLA lag screw

3.2.2. Other bioabsorbable implants

In the clinical study dealing with intra-articular femoral head fractures associated with posterior luxation of the hip, two patients were treated with fully threaded SR-PLLA screws 4.5 mm in outer and 3.2 mm in inner diameter. The lengths of the screws were 45-70 mm. The bending strengths were 200-250 MPa, shear strength 110 MPa, and bending modulus 5-6 GPa.

In four cases, SR-PLLA rods 2 mm in diameter and 25-50 mm in the length were used. In one of these four patients, one SR-PGA rod 2 mm in diameter and 30 mm in the length was used additionally. The SR-PLLA rods had the following strength values after gamma-radiation: the bending strength 200-300 MPa, shear strength

100-180 MPa, and bending modulus 6 GPa. The corresponding values of SR-PGA rods after ethylene oxide sterilization were 220-400 MPa, 180-250 MPa, and 10-15 GPa.

3.2.3. Metallic screws

In paper I, when comparing bioabsorbable and standard metallic cancellous bone screws, screws with a nominal outer diameter of 6.5 mm and an inner diameter of 4.5 mm were used in the control group. In a clinical study of femoral neck fractures (IV), metallic fixations in the control group were performed by using Mecron screws or Ullevaal Hip Screws. The nominal inner diameter of the Mecron screws was 5 mm and the outer diameter 7 mm. The diameter of the shank of Ullevaal Hip Screws was 7 mm, as was also the outer diameter of the threads.

3.3. EXPERIMENTAL STUDIES

3.3.1. Material and methods

3.3.1.1. Surgical technique

The Ethical Committee for Animal Experiments of Helsinki University Central Hospital approved the plan of the present experimental studies before starting the investigations.

53 Finnish young adult sheep of either sex were operated on for these series. The mean weight of the sheep was 53.2 kg (range 31-83). The operations were carried out at the Department of Clinical Sciences of the College of Veterinary Medicine in Helsinki later the Faculty of Veterinary Medicine, University of Helsinki. After the operation, the sheep were kept in the big animal stable of the same department except for the sheep with a follow up period of more than 24 months, which were moved two months after the operation to a farm specialized in raising sheep. At the operation time all the sheep were clinically healthy. One sheep died 24 hours after operation and autopsy showed pneumonia to have caused its death.

The sheep were fasting two days before the operation with a free supply of water. As premedication they were given 1 mg atropine (Atropin®, 1 mg/ml, Orion, Espoo, Finland) subcutaneously (s.c.) and were anaesthetized with intramuscular (i.m.) injections of 0.025 mg/kg medetomidine (Domitor®, 1 mg/ml, Lääkefarmos, Turku, Finland) and 1.0 mg /kg ketamine hydrochloride (Ketalar®, 50 mg/ml, Parke Davis, Barcelona, Spain). During the operation the sheep received every 30-60 minutes 50 % of the initial doses of medetomidine and ketamine hydrochloride i.m. The sheep were positioned on the right lateral recumbency, and the left hip was shaved and scrubbed with polyvidon-ionide (Betadine®, 100 mg/ml, Leiras, Tammisaari, Finland) and chlorhexidine gluconate (Klorhexol®, 5 mg/ml, Leiras, Tammisaari, Finland) antiseptic solution. A direct lateral exposure was analogous to the approach used in orthopaedics (Hardinge 1982). The femoral neck was exposed and a complete transverse subcapital osteotomy was performed with an oscillating saw in 24 sheep (papers I and II). The osteotomy was reduced and fixed temporary with clamps. Two channels parallel to the femoral neck were drilled with a 3.5 drill bit through the osteotomy from the lateral cortex just beneath the greater trochanter to the subchondral bone of the femoral head. The drill channel was measured, and the tapping was performed. The countersinking was made and, after inserting the SR-PLLA screws 4.5 mm in outer diameter, the excess of the screws heads were cut off with an oscillating saw 1 mm above the level of the cortex maintaining the lag screw effect (Fig.4).

Figure 4. Schematic drawing of a subcapital femoral osteotomy in a sheep fixed with two SR-PLLA lag screws.

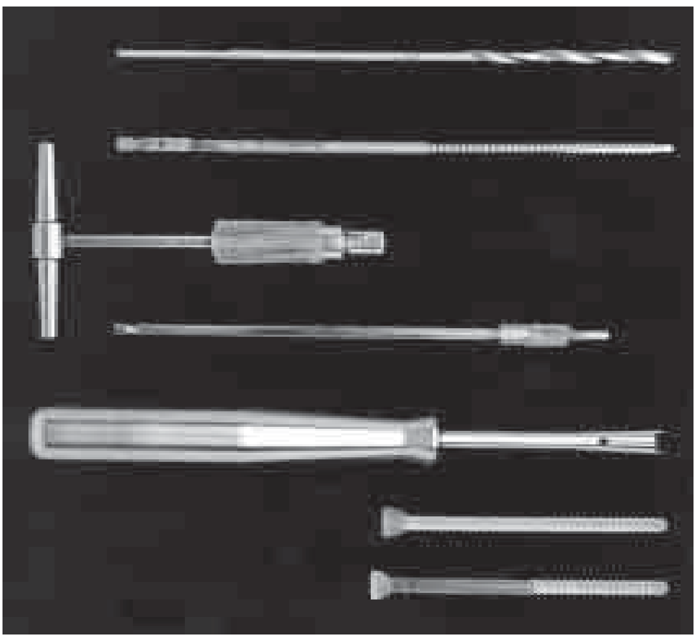


Figure 5. Special instruments used for SR-PLLA screws. From the top: drill bit, tapping instrument, T-handle, countersinking instrument, screw driver, SR-PLLA screw and SR-PLLA lag screw.

A special tapping instrument, countersink and screwdriver developed for the absorbable implants were used (Fig.5). The joint capsule was closed, the middle gluteal muscle was reattached to the greater trochanter and the wound was closed in layers with absorbable sutures (Vicryl®, Ethicon, Nordenstedt, Germany).

In 29 sheep, the lateral approach to the upper femur was used as above except for the opening of the capsule. Without any osteotomy, one SR-PLLA screw 6.3 mm in outer diameter was inserted in the way described above. A 4.5 mm drill bit was used. In 20 sheep the countersinking was not made and the excess part of the screw head was left to get a grip in the pull-out test. Five of these sheep operated on in the pull out test series were used for the subcutaneous implantation of screws. Four SR-PLLA lag-screws 4.5 mm in outer diameter and four SR-PLLA lag-screws of 6.3.mm in diameter were implanted in the subcutaneous tissue of the left side of the chest. The operation was performed in the same anaesthesia after shaving and scrubbing through a small incision, which was closed with Vicryl® sutures.

Before recovery from anaesthesia the sheep were given phenylbutazone 20 mg/kg (Reumuzol®, 200 mg/ml, Lääkefarmos) to kill pain and 60 000 IU/ kg benzylpenicillin procain (Procain Penicillin.G ®, Orion, Espoo, Finland) for prophylaxis of infection. This medication was continued 3 days postoperatively. After the operation the sheep were allowed to move freely without any external support. Normal ruminant food and water were given.

3.3.1.2 Examination methods

3.3.1.2.1 radiologocal methods

Three weeks after the operation, the sheep with an osteonomy were anaesthetized and anteroposterior radiographs of the pelvis were taken to check possible dislocations of failing of the fixations of the osteotomies (Siemens Polyphos 30 M). After the follow-up periods the sheep were sacrified, and both upper femurs were removed and the soft tissue dissected. Radiographs were taken in the anteroposterior position of all specimens except those used for the pull-out test. Kodak T-MAT G/ RA film (Rochester NY) was used with technical values of 40 kV, 8 mAs, 0.04 s exposure time and 100 cm distance. In the radiographs, reduction and dislocation, external callus formation and the union rate of the osteotomy were analyzed as well as the bone density in the screw area and in the surrounding bone.

In three specimens with a follow-up time of five years and in one of seven years and four months, CT and MRI examinations were performed.

With the General Electric Hispeed Advantage computed tomography system (GE Medical System, Milwaukee, USA), contiguous one-mm-thick slices parallel to the long axis of the neck of femur were obtained of both hips before dissection. The CT-density, i.e. Hounsfield unit (HU), was measured on ten separate regions of interest (ROI) in the implant area (drill channel) and also of ten ROI in the surrounding cancellous bone as well as on the contralateral hip. The density of the cortical bone was also assessed in the same manner on both hips. The mean values and the standard deviations (SD) were calculated. The CT density value, named after Hounsfield, is an arbitrary unit and represents a relative linear absorption coefficient. By means of internal calibration of the scanners, the density value of water is set at 0 and that of air at 1000. The absorption values of other tissues are expressed in relation to this scale.

In three specimens with a follow-up time of five years, the MRI was performed on an open 0.23 T MR unit (Outlook, Picker Nordstar Inc, Vantaa, Finland). T1 weighted (TR 360, TE 24) Spin echo (SE) images with three-mm-slice thickness were obtained parallel to the long axis of the femoral neck. The imaging time was 5 min and 11 s with two averaging. FA was 90 degrees and FOV was 169 mm x 200 mm. In one sheep sacrified seven years and four months after the operation, MRI was performed before dissection with a 1.5 T high-field magnet (Magnetom Vision, Siemens, Erlangen, Germany) by means of a flexible local coil. A coronal T1 sequence (TR/TE 600/20 ms) with two-mm-slice thickness and 512 matrix was obtained.

3.3.1.2.2. Histological methods

For the histological examination of the tissue response to PLLA in the long-term (II and III) both proximal femurs of 19 sheep were, after dissection, fixed in a series of ethanol immersions of increasing concentrations and embedded in methylmetachrylate (Schenk 1965). Sagittal sections five micrometers thick were cut with a microtome (Polycut S, Reichert-Jung, Nussloch, Germany). The sections were stained using the Masson-Goldner trichrome method (Goldner 1938). A Leitz Diaplan microscopy (Leitz, Wetzlar, Germany) was used for the qualitative histological evaluation. With the means of polarising microscopy the birefrigent polymeric material could be identified in the sections.

3.3.1.2.3. Microradiography

For the microradiography thicker sections measuring 80 micrometers were cut from the same blocks with a Leitz Saw Microtomy 1600 (Wetzlar, Germany). High Resolution Ultra Flat Plates Type 1 A (Microchrome Technology Inc.) were used in the Faxitron x-ray system (Hewlett Packard, McMinnville, Oregon) for the contact microradiographs taken with 21 kV and an exposure time of 15 minutes.

3.3.1.2.4. Strength measurements of load carrying capacity of osteotomized bones fixed with SR-PLLA screws and metallic screws.

Twelve weeks after the operation the sheep were sacrificed. After dissection and radiographic examination the metallic screws were removed, whereas the SR-PLLA screws were left in place. The specimens were transferred into 0.9% saline solution, maintained at room temperature (22-23° C) and send immediatly to Tampere University of Technology, where all mechanical testing was carried out within 24 hours after the sacrificing of sheep. A materials testing machine (model 6000R, Lloyd Instruments PLC, Fareham, U.K.) interfaced with an IBM PS/2 386 computer using Lloyd Series R software for control of operation and data acquisition was used. For testing purposes, the femoral diaphysis was embedded in an aluminium can section, using Acryfix 5Q cold mounting resin (Streurs, Rodovre, Copenhagen, Denmark) and was clamped distally in a machinist´s adjustable vice with the diaphyseal axis placed vertically (Fig. 6).

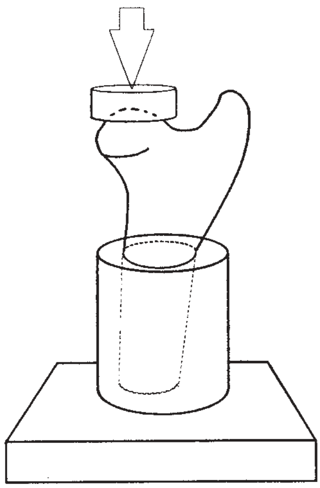


Figure 6. Schematic representation of the biomechanical testing set up for the loading bone specimens mounted into Lloyd 6000R Materials Testing Machine..

All experiments were conducted at room temperature, and moisture conditions were controlled by spraying the specimens with saline and wrapping them in paper towels. Load was applied to the femoral head vertically through a polyethylene cup at a rate of 10 mm/min. The specimens were tested to failure. The load at failure (expressed in N) was determined from load-displacement data as the maximum load before fracture. The intact unoperated right femur of each sheep acted as a control.

3.3.1.2.5. Strength measurements of the SR-PLLA lag screws

The SR-PLLA lag screws were removed carefully from the surrounding subcutaneous tissue of the sheep and transferred to saline solution and sent for testing within 24 hours. In addition intact sterilized lag screws were tested.

The bending strength was measured by the three-point bending method (Fig. 7), using a Lloyd LR 30K Materials Testing Machine (Lloyd Instruments PLC, Fareham, U.K.).

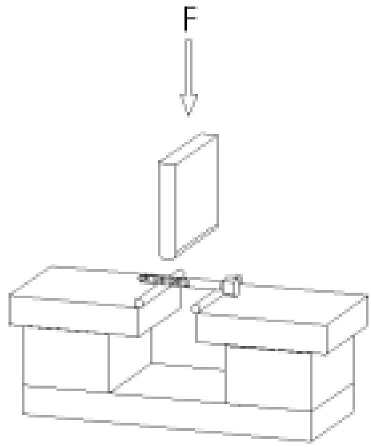


Figure 7. Schematic drawing of the three-point bending tool for measurement of the bending strength of SR-PLLA screws.

At least three samples of each screw size and follow-up period were tested. As drying of the incubated screws can lead to a decrease in their strength, the measurements were performed on wet samples. The support spans and the cross-head speeds were 42 mm and 10 mm/min. The radius of the loading nose was 5 mm and radius of each support was 1.5 mm. The bending strength was calculated using the following equalition:

$$\sigma_b = \frac{8 \cdot F_{max} \cdot L}{\pi \cdot d^3}$$

where σ = bending strength (MPa)
F_{max} = maximum bending force (N)
L = support span (mm)
d = inner diameter of the screw

The shear strengths of the SR-PLLA screws when intact and after in vivo exposure were measured by means of a tool, which was constructed by modifying the standard BS (British Standard) 2782, Method 340 B (1978). The tool consisted of two parts, which were joined together by the implant (Fig. 8 a). During the test the parts were pulled apart, using a Lloyd LR 30K Materials Testing Machine operating at a cross-head speed of 10 mm/min, such that the implant resting in a drillhole was cut into pieces perpendicularly to the long axis of the screw (Fig. 8 b).

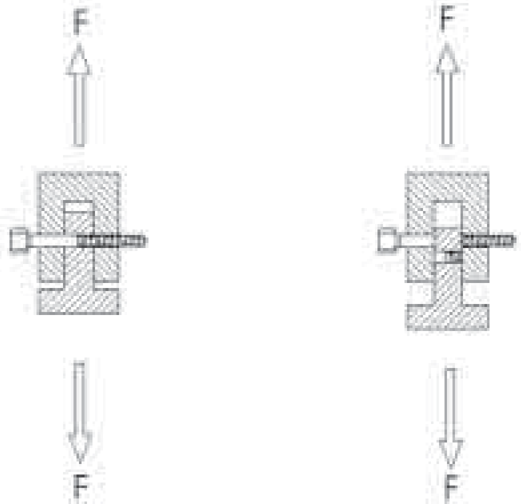


Figure 8 a-b. The testing arrangements for measuring the shear strength of SR-PLLA screws before (a) and after the test (b).

The shear strength was calculated using the following equation:

$$\tau_s = \frac{2 \cdot F_{max}}{\pi \cdot d^2}$$

where τ_s = shear strength (MPa)
F_{max} = maximum force recorded (N)
d = inner diameter of the screw (mm)

For testing the pull-out strength of SR-PLLA lag-screws the femurs were embedded in Acryfix SQ (Struers, Rodovre, Copenhagen, Denmark) cold mounting resin and clamped in a machinist´s vise with the diaphyseal axis placed at approximately 45 degrees angle. Tensile load was applied by means of a wire rope anchored to the screw head (Fig. 9). Pull-out load was applied parallel to the screw axis at a rate of 10 mm/min. The load at failure (expressed in Newtons) was determined from load-displacement data as the maximum force to break or pull out the screw.

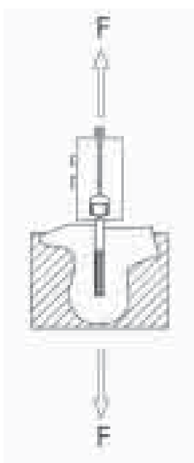


Figure 9. Schematic drawing of the bone specimen mounted for pull-out testing of inserted SR-PLLA

The arrangement of the tensile strength testing was similar to the pull out test except that the screw was turned into the jaw instead of bone (Fig 10).

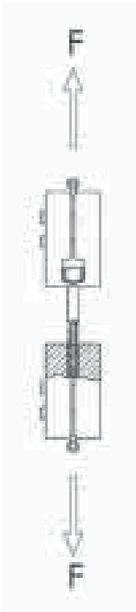


Figure 10. The testing arrangement for the tensile strength measurement of SR-PLLA screws.

3.3.1.2.6. Molecular weight measurements

The solution viscosities (according to ASTM, American Society for Testing and Materials, D 445-88) of intact SR-PLLA screws and after in vitro and in vivo exposure were measured in chloroform at 25° C with an Ubbelohde capillary viscometer (type Oa according to ASTM D 446). The intrinsic viscosities (X in dl/g) were obtained by linear regression analysis from the dilution series (0.1 g/dl, 0.2 g/dl, 0.3 g/dl and 0.5 g/dl) and viscosity-average molecular weights (in g/mol) were calculated using the Mark-Houwink equation and the parameters were determined by Schindler and Harper (1979):

$$[\eta] = 5.45 \cdot 10^{-4} \cdot M_v^{\,0.73}$$

3.4. CLINICAL STUDIES

3.4.1. General remarks

The present study, comprised of two clinical series, includes 86 patients. The operations were performed between August 1989 and October 1994 at the Department of Orthopaedics and Traumatology of the Helsinki University Central Hospital. The study was performed with permission of the Ethical Committee of the Department.

3.4.2. Subcapital femoral neck fractures

3.4.2.1. Patients and methods

Between January 1991 and October 1994, 40 patients, 34 females and six males, with a subcapital femoral neck fracture were treated operatively by using three bioabsorbable SR-PLLA screws 6.3 mm in outer diameter. The mean age of the patients was 73 years (range 36-96 years) and the mean weight 61 kg (range 40-90 kg). Thirty-three patients were operated on within two days and all within 26 days after injury. Postoperatively, the patients were allowed to walk on crutches and partial weight-bearing was recommended if possible. Full weight-bearing was allowed after six weeks. The patients were followed up prospectively and they visited the outpatient department for clinical and radiological check-up at six and 12 weeks, and at six, 12, 24 and 36 months postoperatively and later if necessary. In the case of redisplacement, the follow-up time given in this study ended at the time of verification of the failure of the fixation.

In the same period of time 40 patients with a subcapital femoral neck fracture were operated on, using metallic Mecron or Ullevaal Hip Screws. The distribution in the Garden classification was the same as in the SR-PLLA-group (29 Garden I-II, nine Garden III and two Garden IV fractures). There were 26 females and 14 males in this group, the mean age of the patients being 70 years (range 38-88 years) and the weight 65 kg on average (range 45-87 kg). The operations of 33 patients were carried out within two days and all within 26 days. Postoperatively, most of the patients were allowed to walk on crutches with partial weight-bearing, though some patients were permitted weight-bearing as tolerated from the first postoperative day. The patients with metallic fixation visited the outpatient department for check-up at six weeks and three, six and 12 months postoperatively and later on if necessary. The follow-up time given here ended if verification of the redisplacement was set.

All the patients were operated on in spinal anaesthesia. The displacement was reduced and the position was controlled by using an imaging intensifier with the patients on the orthopaedic operating table. From a short lateral approach through the vastus lateralis muscle, the lateral cortex of the femur just beneath of the greater trochanter was exposed. In the bioabsorbable fixation, special, not cannulated, vitallium drills were used to avoid any disturbances caused by small metallic particles in the later MRI control. One K-wire was drilled from the lateral cortex into the middle of the femoral neck at about 130° angle in the imaging intensifier control on two planes to mark the direction of the screws. Three drill holes of 4.5 mm in diameter were drilled parallel to the K-wire in the imaging intensifier control and all bits were left in place. The K-wire was removed. One by one the drills were removed, the lengths of the drill channels were measured and after tapping and countersinking the SR-PLLA screws were inserted. The excess parts of the screw heads were cut off 1 mm above the cortex with a small oscillating saw (Fig.11). The wound was closed in layers. Due to the difference in thread geometry and in the shape of the screw head compared to standard metallic screws, special tapping devices, a screwdriver as well as a special countersink, were needed for the absorbable SR-PLLA screws.

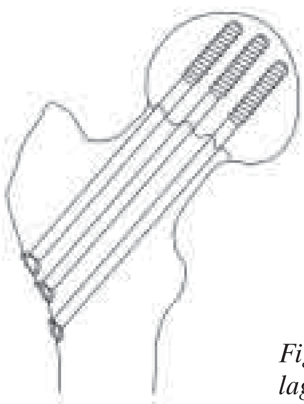


Figure 11. Schematic representation of bioabsorbable lag screw fixation of a femoral neck fracture.

When using metallic screws the standard operative techniques given for each implant were followed. In 38 patients three metallic screws were used for each fixation, in one case, two, and, in another case, four Mecron screws were used.

3.4.3. Femoral head fracture associated with traumatic dislocation of the hip joint

3.4.3.1. Patients and methods

From August 1989 to November 1992, six patients, (five males and one female with a fracture of the femoral head caused by a traumatic posterior dislocation of the hip joint) were treated operatively at our department. Their mean age was 33.2 years (range 20-61) and their average weight was 88 kg (range 55-126). Four patients had sustained a traffic accident, one had fallen in downhill skiing and one patient had been injured at work by caving in of the ground. According to the Pipkin classification (1957) there were one of type 1, four of type 2 and one of type 4 fractures. In five patients there was only one main fragment with or without smaller fragmentation, while in one patient there were three fragments (Fig. 12). The size of the fragments varied from one-fifth to three-fourths of the cartilage surface of the head. The open reduction and internal fixation were performed with self-reinforced bioabsorbable implants.

The dislocations of the hip joint were first managed by emergency closed reduction with the exception of one patient in whom the diagnosis was set with a delay of 24 days. The latter patient was multiply injured and brought to our hospital from abroad after operative treatment of other injuries. In two cases the initial closed reduction failed. All the patients were treated by internal fixation. Four patients were operated on within 24 hours, one patient was transported from another hospital and was operated on three days after the injury. The patient from abroad was operated on immediately after setting the diagnosis. Five hips were managed through a posterior and one through a lateral approach. T-shaped capsulotomy was performed if necessary. The fragments were reduced under direct vision and fixed with bioabsorbable SR-PLLA or SR-PGA implants. In four patients smaller fragments were fixed using bioabsorbable rods through the cartilage surface. The drill channel, 2 or 3.2 mm in diameter, was drilled and after measurement a rod of the same diameter was inserted with a special applicator. The base of the implant was sunk 1-2 mm under the cartilage surface.

Figure 12. A preoperative view of the femoral head fracture fixed with bioabsorbable rods. The rods are left above the cartilage surface for the photograph. Finally they were sunk 1 mm below the surface.

Figure 13. Schematic representation of the fixation of the fragment in Pipkin type 2 femoral head fracture by two SR-PLLA rods.

In two patients with bigger fragments the fixation was managed extra-articularly with two SR-PLLA screws. After an open transarticular reduction the drill channel was drilled from the femoral neck to the subchondral bone of the head fragment. The tapping and countersinking were performed and after insertion of the screw, the excess of the screw head was removed with an oscillating saw. The mean operation time was 1 h 57 min (range 1 h 10 min -3 h 15 min). The estimated blood loss was 530 ml (range 200-1100 ml) on average.

From the first postoperative day five of the patients were allowed to walk around on crutches with gradually added partial weight bearing for 5-12 weeks (seven weeks on average). Full weight bearing was permitted after six to 14 weeks postoperatively (mean 10 weeks). One patient with multiple injuries was treated by bed rest. This patient died six weeks later, mainly because of cerebral trauma.

3.4.4. Statistical methods

T-tests and multivariate analysis were used for statistical analysis. In the study of subcapital femoral fractures (IV), the chi-square test with Yates´ correction was applied to the ability to walk and to the range of motion due to two-sided hypothesis. Pain was tested with the Wilcoxon rank test.

4. RESULTS

4.1. COMPARISON OF THE FIXATION OF SUBCAPITAL FEMORAL NECK OSTEOTOMIES WITH ABSORBABLE SELF-REINFORCED POLY-L-LACTIDE LAG-SCREWS OR METALLIC SCREWS IN SHEEP (I)

4.1.1.Radiological results

Three weeks. In the sheep with subcapital osteotomy anteroposterior radiographs were taken at three weeks. Failure of fixation was observed in one out of seven sheep treated with SR-PLLA lag-screws as well one out of seven sheep with metallic fixation. These sheep were killed after the examination. All other osteotomies maintained their positions.

Twelve weeks. The radiographic analysis after sacrificing at 12 weeks showed a bony union in all the osteotomies (Fig. 14 a-b). Callus formation was seen in three sheep in the SR-PLLA as well as in three sheep in the metallic group. There was one insignificant primary redisplacement in each group and one insignificant secondary redisplacement occurred after metallic fixation.

Figure 14 a-b. Anteroposterior radiographs of left femurs of sheep 12 weeks after aubcapital osteotomy fixed with two SR-PLLA lag screws (a) or two metallic cancellous bone screws (b).

4.1.2. Strength measurements

There were no statistical differences between the groups in respect to the load needed to break the osteotomies fixed with SR-PLLA or metallic screws. The mean load at failure in the SR-PLLA group was 5576.6 N (range 2525-6892 N) and in the metallic group 4448.8 N (range 3175-8434 N). When compared with the average load-carrying capacity of the control specimens, the average load at failure of the operated bones was 94.7% in the SR-PLLA group and 88.3% in the metallic group. In the group treated with bioabsorbable screws, the control femoral necks were 117% stronger on average than those in the metallic group. No statistically significant differences were found between the SR-PLLA and metallic screw groups. Typical load-displacement curves for selected specimens fixed with metallic screws or SR-PLLA screws as well as for control bone are shown in Fig. 15.

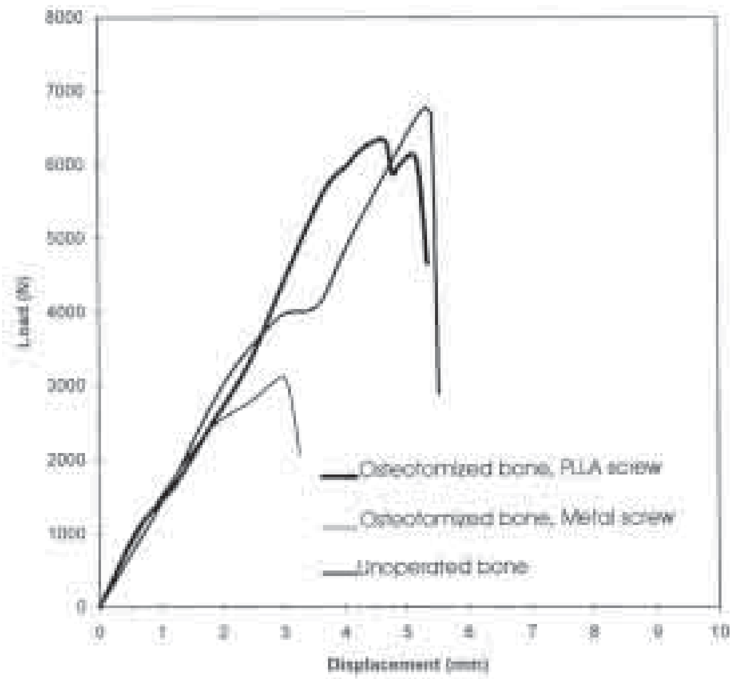


Figure 15. Load (N) displacement (mm) curves of selected samples. Note: Metallic screws were removed prior to biomechanical testing of the bones, whereas SR-PLLA screws were left in place.

4.2. HEALING OF SUBCAPITAL OSTEOTOMIES FIXED WITH SELF-REINFORCED POLY-L-LACTIDE SCREWS; AN EXPERIMENTAL LONG-TERM STUDY IN SHEEP (II)

4.2.1. Radiological results

Three weeks. In one sheep quite a strong ectopic bone formation was visible, but no redislocation and no abnormal clinical signs.

Twelve weeks. All three osteotomies had united. In two specimens the upper part of the osteotomy line was still visible.

One year. All the osteotomies fixed with SR-PLLA screws had healed. In one specimen there was ectopic bone formation at the site of the neck and greater trochanter but not redislocation. The bone density had decreased (Fig. 16).

Three years. The osteotomies showed a bony union and the sites of the screws were visible, but the density of the bone in the screw area was almost the same as in the surrounding bone and in the unoperated right femur. In one specimen there was strong exostosis formation around the whole upper part of the femur and acetabulum, which was already observed in the control radiographs at three weeks.

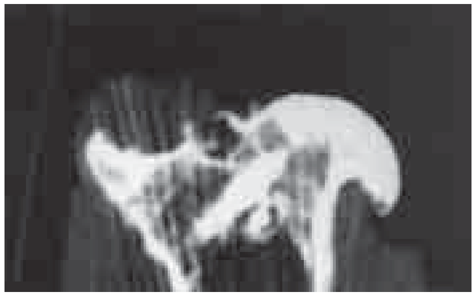
Seven years and four months. In one sheep killed seven years and four months after the osteotomy was fixed with two SR-PLLA screws, the osteotomy was united without any dislocation, though the head was flattened down to some extent. The bone density was increased at the site of the implant as well as in the femoral head (Fig.17a).

The CT-density values were higher in the screw area (mean HU 2226, SD 205) than in the surrounding cancellous bone (mean 614 HU, SD 118) or in the cancellous bone in the contralateral hip (mean HU 482) corresponding to the values obtained from the compact cortical bone (mean HU 2212, SD 123)(Fig.17b).

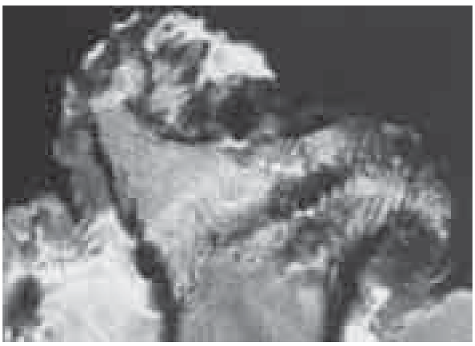
In the MRI the site of the implant was visible as a channel of reduced signal intensity, reflecting sclerotic bone and fibrous tissue. The channel was surrounded by a patchy rim of bone with a slightly increased T1 - signal corresponding to yellow bone marrow with signs of fatty conversion. No other changes were observed (Fig. 17c).

Figure 16. Anteroposterior radiograph of the left femur of a sheep one year after a subcapital osteotomy fixed with two SR-PLLA screws. The osteotomy has united but the screw channels are still visible, showing a decreased bone density with increased density at the borders.

A



B



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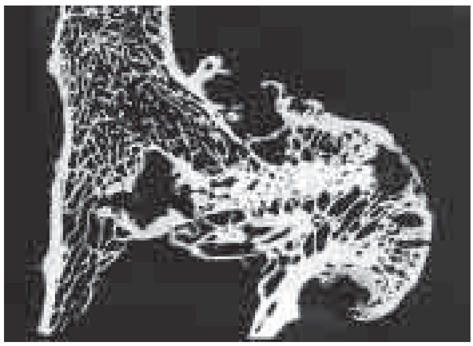


D

Figure 17 a-d. Radiography (a), CT scan (b), MRI (c), and microradiography (d) of the proximal femur of the sheep 7 years and 4 moths after an SR-PLLA screw fixed subcapital osteotomy. Strong bone formation is seen at the site of the implant, where the bone density measured on ct was at the level of cortical bone. On MRI the screw area was visible as a channel of reduced intensity, reflecting sclerotic bone surrounded by a patchy rim of bone.

4.2.2. Histological results

Twelve weeks. At twelve weeks after subcapital osteotomy fixed with two SR-PLLA lag screws, a bony union of osteotomies, partly with cartilaginous healing was observed. There was new bone formation with osteoblasts and osteoid seen at the tissue-implant boundary. Connective tissue was seen as a narrow layer around the screws. In two specimens no tissue response or signs of inflammation were seen. A thin layer of connective tissue with some giant cells surrounded the implant. In one specimen there was a small focal area with some lymphocytes and macrophages around the superficially loosened fibres of the implant. The structure of the implant looked generally still intact (Fig. 18).

One year. The bony union and new bone formation with osteoblasts and osteoid outlining the implants were observed. At the tissue-implant boundary there was a mild foreign-body reaction - a narrow layer of connective tissue with some giant cells surrounding the implants. The connective tissue started to penetrate into the implant from the core area. There was a mild foreign-body reaction, with a few lymphocytes and foam like macrophages (Fig. 19).

Three years. A mild or moderate foreign-body reaction was observed. The implant was split into smaller fragments, which were surrounded by connective tissue and around the bigger fragments some lymphocytes and polymorphonuclear macrophages were also visible (Fig. 20). Polarized light showed small islands of birefringent material of the implant.

Seven years and four months. Polarized light did not show any birefringent material of the implant even when controlled in 80 μm thick sections cut for the microradiographs (Fig. 21).

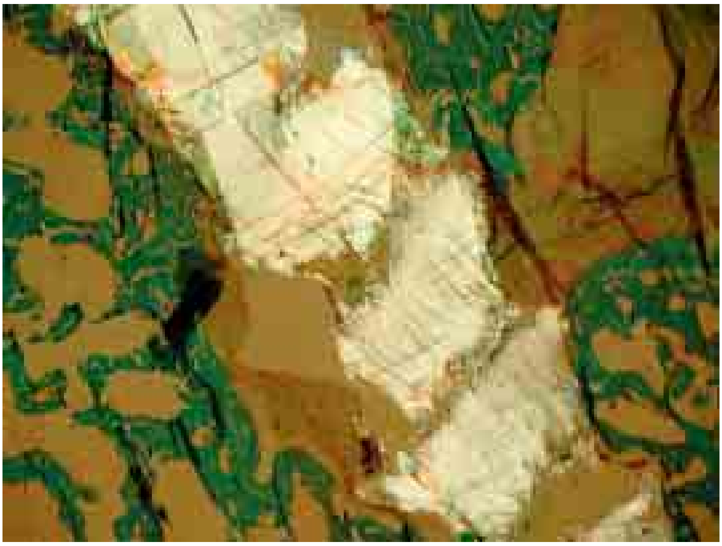


Figure 18. At 12 weeks SR-PLLA screw in situ within the bone. (Masson-Goldner trichrome, polarized light, original magnification x 2).

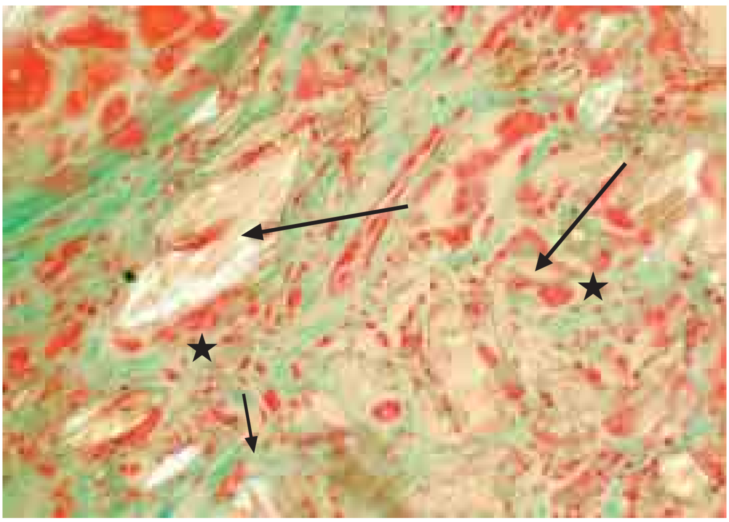


Figure 19. At one year disintegrating SR-PLLA particles (big arrow) surrounded by giant cells (asterisk) and immature connective tissue with plump fibrocyte nuclei (small arrow). (Masson Goldner trichrome, polarized light, original magnification x 20).

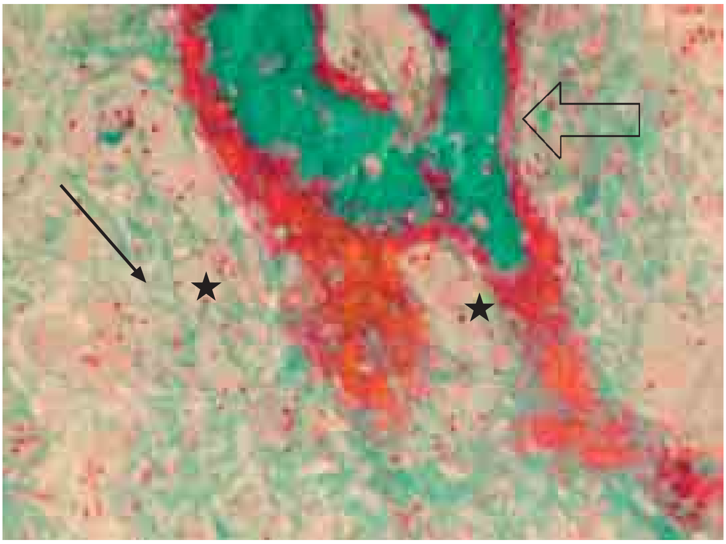


Figure 20. At three years, remnants of SR-PLLA (asterisk) ingested by macrophages. Basophilic connective tissue (arrow) in between. Trabeculae of bone with osteoid rim (open arrow) in the upper part of the picture. (Masson goldner trichrome, original magnification x 20).

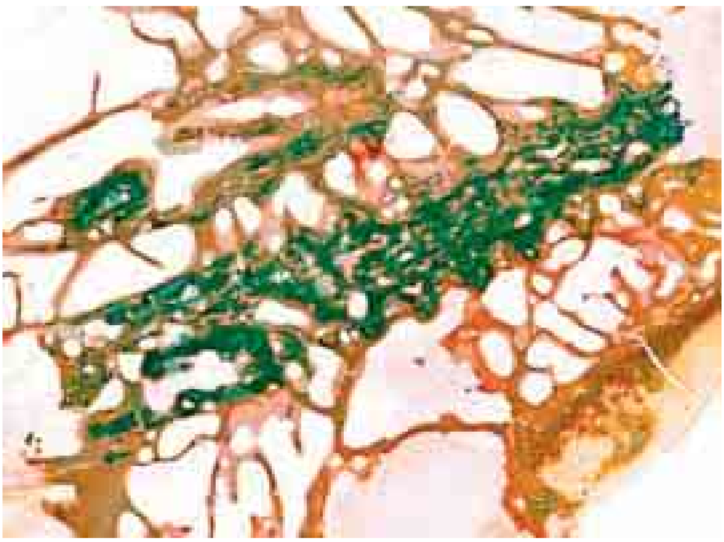


Figure 21. An 80 μm section of a proximal osteotomized femur of a sheep at seven years and four months after SR-PLLA fixation. Screw area is replaced by bone (dark green area). (Masson Goldner trichrome).

4.2.3. Microradiographic results

In the microradiographs at 12 weeks the osteotomies had healed and a strong bony union was visible in every specimen at later follow-up times. Especially around the screws, strong bone trabeculation was observed already at 12 weeks, which increased gradually and at seven years and four months the bone density in the screw area had increased to the level of cortical bone (17d).

4.3. BIODEGRADATION AND STRENGTH RETENTION OF POLY-L-LACTIDE SCREWS IN VIVO. AN EXPERIMENTAL LONG-TERM STUDY IN SHEEP (III)

4.3.1. Radiological results

Two years. In the radiographs at two years after inserting an SR-PLLA screw 6.3 mm in diameter in the proximal femur the bone density had increased at the implant-tissue boundary but in the area of the screw channel it had decreased slightly, compared to the surrounding bone (Fig. 22 a).

Three years. The density in the screw area had increased and there was a sclerotic area around the screws three years after SR-PLLA screw implantation (Fig. 22 b).

Five years. In the plain radiographs five years after the insertion of the SR-PLLA lag-screw the boundaries of the screw area were not easily detectable and the bone density was increased in the whole screw area (Fig. 22 c). In CT examination the density values of the implant area were higher (mean 985 HU, SD 89) that those of the surrounding (mean 507 HU, SD 87) or contralateral (mean 640 HU, SD 74) cancellous bone (Fig. 23). In MRI the drill channel was still visible as a high signal intensity zone surrounded by low signal intensity borders on the T1-weighted SE sequence. Otherwise the bone marrow signal intensity was normal on both sides (Fig. 24).

A

B

C

Figure 22 a-c. Anteroposterior radiographs of the femur of a sheep after implantation of an SR-PLLA lag screw. At two years the screw area had decreased density except at the borders, where the density was increased (a). At three years the density has increased (b) and at five years still more clearly (c).

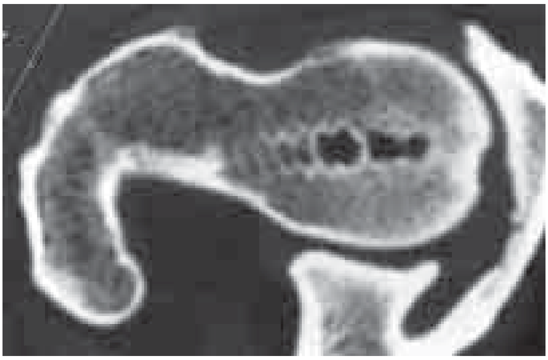


Figure 23. Computed tomography scan of the upper femur of the sheep at five years after implantation of an SR-PLLA screw. CT density of the bone in the implant area was increased in comparison to that of the surrounding cancellous bone.

Figure 24. T1 weighted coronal magnetic resonance imaging five years after implantation of an SR-PLLA screw. The screw area was heterogeneous, consisting of low signal intensity corresponding to bone ingrowth and areas of moderate signal intensity reflecting the fatty bone marrow.

4.3.2. Histological results

Two years. The PLLA implant inserted in the proximal femur was fragmented and surrounded by some polymorphonuclear cells and connective tissue. At the boundaries of the screw area there was strong bone formation. In polarized light the remaining implant was visible as birefrigent material and there were some minor fragments seen outside the original screw area.

Three years. Small islands of implant were seen in the screw area. The small fragments were surrounded by connective tissue and lymphocytes and polymorphonuclear macrophages showing a moderate foreign-body reaction. New bone formation and osteoid were also detectable at the screw area. A strong trabecular bone was outlining the implant profile. In one specimen the birefringent lactide material could not be found in polarized light (Fig. 25 a-b).

Five years. The screw area was surrounded by tight bone formation. There were no signs of a foreign body reaction. Lactide material was not found as birefringent material when examined with polarized light (Fig. 26).

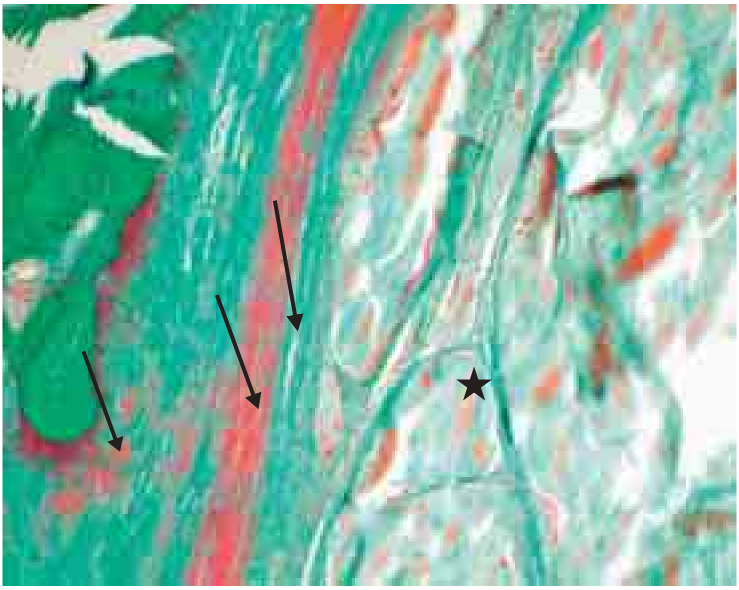


Figure 25 a. At 3 years, disintegrating SR-PLLA (asterisk) with strands of connective tissue (arrow) in between. On the left, mature bone trabeculae with an osteoid rim. (Masson-Goldner trichrome, original magnification x 20).

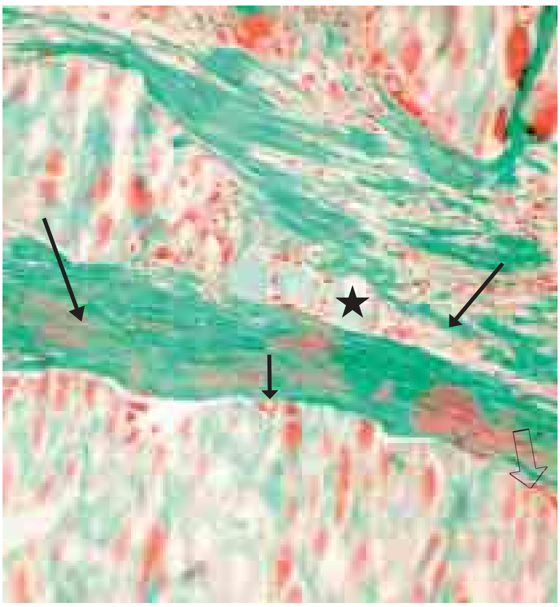


Figure 25 b. At three years, disintegrating SR-PLLA (asterisk) with strands of connective tissue (arrow). Small fibrocytic nuclei in rather mature connective tissue. Some macrophages (small arrow) and a few giant cells (open arrow) are seen. (Masson-Goldner trichrome, original magnification x 20).

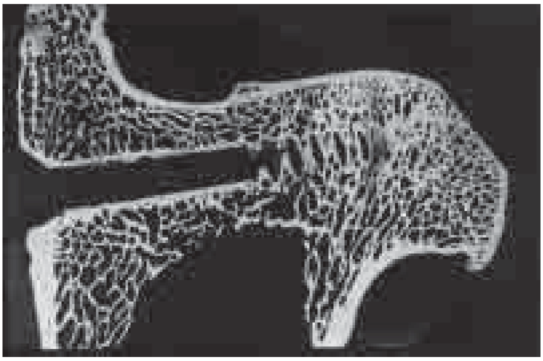
Figure 26. At five years, mature bone has replaced the SR-PLLA screw. Dark green colour indicates the 80 μm thickness of the section. A Haversian channel is seen in the upper left part. (Masson-Goldner trichrome, original magnification x 10).

4.3.3. Microradiographic results

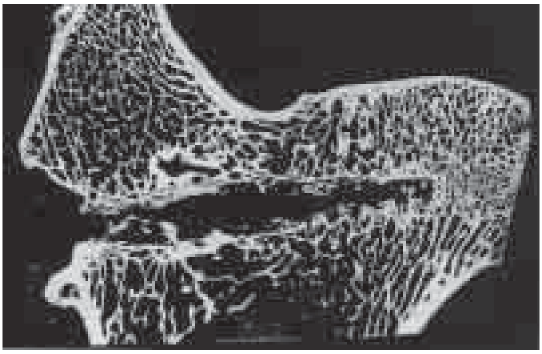
Two years. At two years after insertion of the SR-PLLA lag screws there was no bone formation in the screw area but the bone density was increased at the bone-implant boundary.

Three years Remarkable bone ingrowth was seen in the implant area, which was surrounded by the high density bone area.

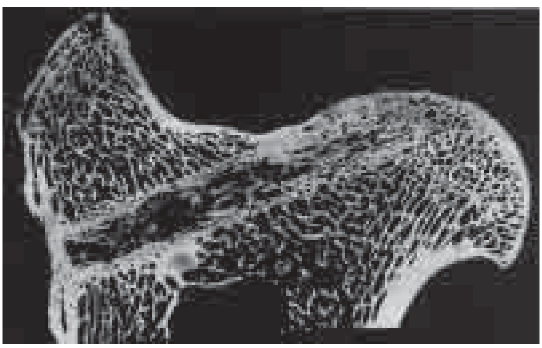
Five years. Strong trabeculae were seen going through the screw area, but the implant area was not completely replaced by bony tissue (Fig. 27 a-c).



A



B



C

Figure 27 a-c. Microradiographs of an 80 μm sagittal section of the proximal femur of the sheep after implantation of an SR-PLLA screw. At two years the bone density has increased in the implant tissue boundary (a). At three years (b) and at five years (c) the screw area showed bone ingrowth.

4.3.4. Strength retention of SR-PLLA screws in vivo

Macroscopic appearance. In the subcutaneous tissue of the left chest the screws might have been exposed to mechanical wear during the implantation when the animals moved or were lying on their left side. Also eight screws implanted from one short incision might have been placed too close to one another, as could be observed while preparing specimens especially at 18 weeks.

The initial bending strength of the 6.3 mm SR-PLLA screws was 257.9 MPa (SD 10.0) and the shear strength 132 MPa (SD 6.4). At 12 weeks the values had decreased to 229.8 MPa and 119.5 MPa, respectively. At 18 weeks the bending strength was 70.3 (SD 36.5), which means 73 % lost of the initial strength. The shear strength was 80.9 (SD 40.5). At 36 weeks the values were 78.8 MPa (SD 23.4) and 55.3 MPa (SD 31.3), respectively. The initial bending strength of 4.5 mm SR-PLLA lag-screws was 176.9 MPa (SD 7.1) and shear strength 162.5 MPa (SD 7.3). At 12 weeks the values were 177.0 MPa (SD 6.0) and 128.0 MPa (SD 10.9). At 18 weeks the bending strength was 147.3 MPa (SD 30.2) and the shear strength 128.1 MPa (SD 1.6) (Fig. 28-29). The bending modulus (GPa) is given in figure 30 and the tensile strength of the lag-screws 6.3 mm in diameter are given in figure 31.

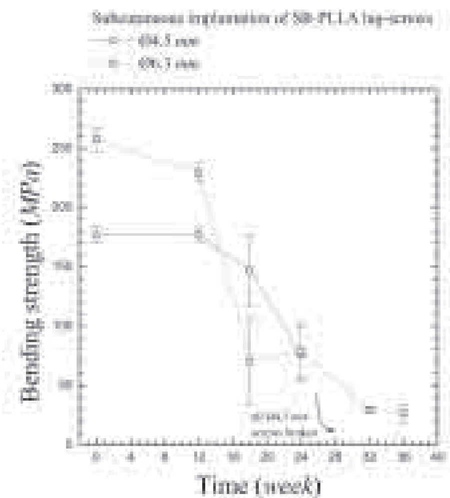


Figure 28. Loss of bending strength of SR-PLLA lag screws in the subcutaneous tissue of the sheep during a 36-week follow-up.

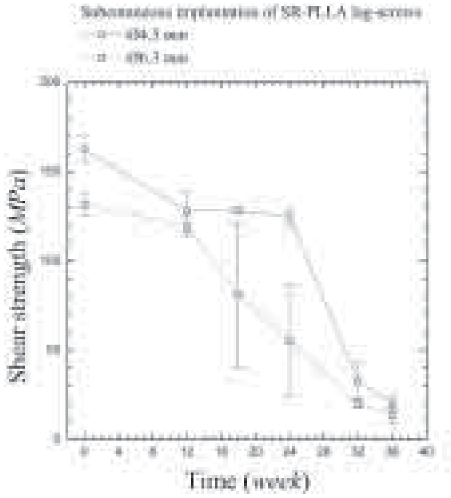


Figure 29. Loss of shear strength of SR-PLLA lag screws in the subcutaneous tissue of the sheep during a 36-week follow-up.

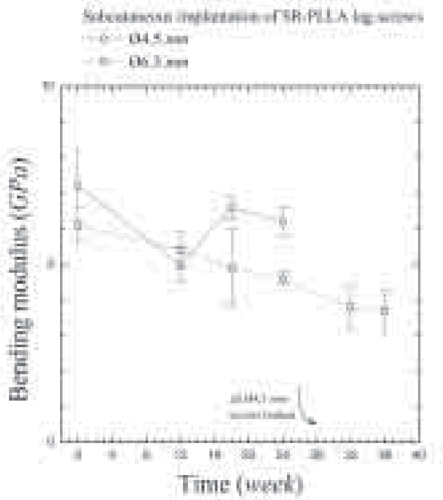


Figure 30. The bending modulus of SR-PLLA lag screws in the subcutaneous tissue of the sheep during a 36-week follow-up.

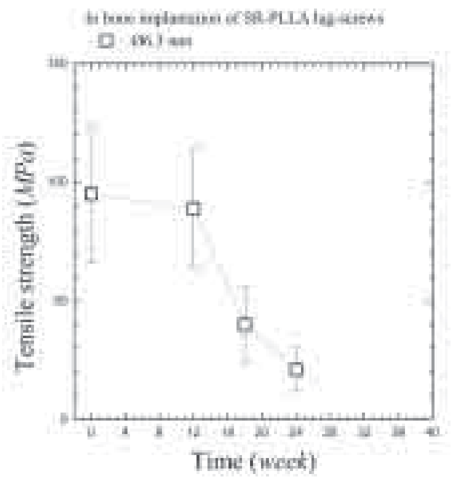


Figure 31. The loss of tensile strength of SR-PLLA lag screws in the bone tissue of the sheep during 36-week follow-up. Measurements were not possible at 32 and 36 weeks because of the mechanical wear of the screws during implantation.

4.3.5. Pull-out tests

The initial pull-out force (the maximum force needed to pull the screw from the bone or until it breaks) was 1507 N (SD 464). The pull-out force decreased by about 6 % of the initial value to 1415N (SD 394) within 12 weeks, nearly 58% to 633 N (SD 247) within 18 weeks, and about 78% to 311 N (SD 137) within 24 weeks (Fig. 32).

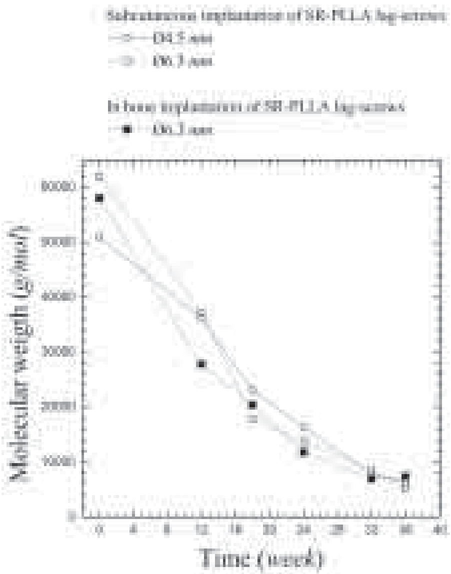


Figure 32. Loss of pull-out force of SR-PLLA screws from the bone tissue. At 32 weeks, three out of four and at 36 weeks two out of four screws were too weak for the measurement, and the head of the screw broke off at the trial.

4.3.6. Molecular weight measurements

The initial viscosity average molecular weight of 6.3 mm pull-out screws implanted in the bone was 58000. At 12 weeks the molecular weight was 48% (27800), at 18 weeks 35% (20400) and at 24 weeks 20 % (11800) of the initial value (Fig. 33).

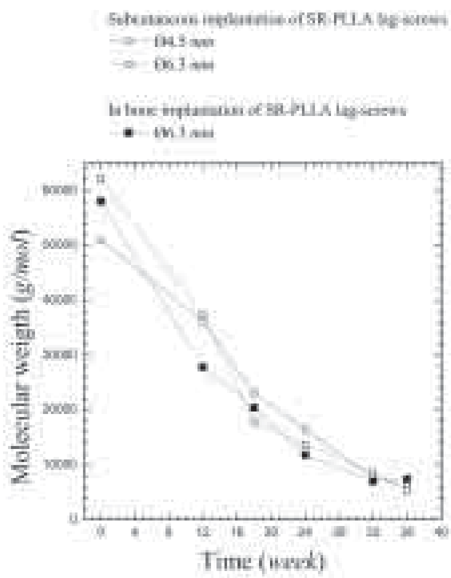


Figure 33. Loss of viscosity average molecular weight of SR-PLLA lag screws in bony tissue and in subcutaneous tissue of sheep during a 36-week follow-up.

4.4. TREATMENT OF SUBCAPITAL FEMORAL NECK FRACTURES WITH BIOABSORBABLE OR METALLIC SCREW FIXATION. A PRELIMINARY REPORT (IV)

The mean follow-up time was 16 months (SD 13.1) for the patients treated with bioabsorbable fixation and 11 months (SD 7.6) for the patients with metallic screws. In Garden stage I and II (impacted or minimally displaced) subcapital femoral neck fractures, there were 5/29 redislocations after SR-PLLA and 8/29 after metallic fixation. In one patient treated with bioabsorbable fixation the fracture united in an acceptable position in spite of a redisplacement of the achieved position. In partially displaced fractures (Garden III) there were 4/9 displacements in both groups. In Garden IV (totally displaced) fractures 2/2 fixations failed. An aseptic femoral head necrosis occurred in one patient treated with SR-PLLA screws after uneventful healing of the fracture. Two further patients with united fractures were examined by MRI because of pain and a transient aseptic necrosis of the head was verified. The pain relieved and the control MRI showed the recovery of the blood circulation. No changes were seen in the plain radiography control. There was also one aseptic necrosis and one transient necrosis after metallic fixation (Table 3).

Table 3. Patients with subcapital femoral neck fractures treated with SR-PLLA or with metallic screws and outcome.

	Garden stage I		**Garden stage II**		**Garden stage III**		**Garden stage IV**	
	SR-PLLA	Metal	SR-PLLA	Metal	SR-PLLA	Metal	SR-PLLA	Metal
Total number	2	2	27	27	9	9	2	2
Sex (female/male)	2/0	2/0	25/2	17/10	5/4	5/4	2/0	2/0
Mean age (years)	91	84	70.5	66	74.5	67	72.5	71.5
Mean weight (kg)	57.5	62.5	60	61.5	68.5	69	63.5	55
Fracture union	2	2	22	19	5	5	0	0
Redislocation	0	0	5	8	4	4	2	2
Aseptic necrosis	0	0	1	1	0	0	0	0
Transient aseptic necrosis	0	0	1	1	1	0	0	0
Removal of screws	0	0	0	8	0	5	0	0

Table 4. Clinical results after 29 SR-PLLA and 25 metallic fixations of femoral head

	SR-PLLA m = 29	Metallic m = 25	
Ability to walk:			
Without difficulty	26	14	
With one crutch or two	3	11	
Not able	0	0	
			P<0.05[1)]
Range of movement:[2)]			
Full ROM	29	14	
Slightly restricted	0	7	
Restricted	0	0	
			P<0.01[3)]
Pain:			
No	25	17	
Slight	3	7	
Difficult	1	1	
			n.s.

[1)]Chi with Yates' correction, [2)]Information not available in for patients, [3)]Wilcoxon rank test

The clinical results are presented in table 4. The patients re-operated on due to a redisplacement or an aseptic necrosis of the femoral head were excluded from the evaluation of the clinical results. The ability to walk was significantly better in the patients treated with SR-PLLA screws ($p < 0.05$), as was also the range of movement ($p < 0.01$), compared to the patients with metallic screw fixation. There was a tendency among the patients treated with bioabsorbable screws to have less pain, though it was not statistically significant.

Thirteen removal procedures of the osteosynthesis devices were carried out 17 (9-25) months after metallic screw fixation. The indications for the operations were pain or mechanical irritation caused by marked backing of the screws (Figs. 34-37).

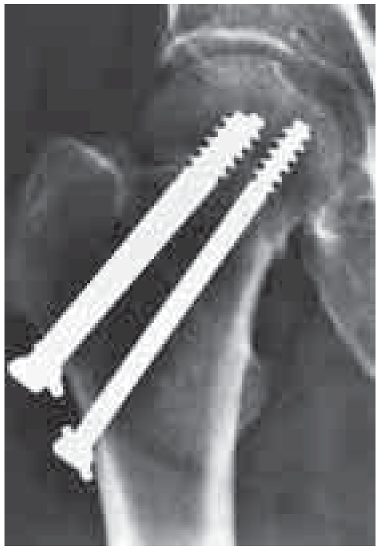
A

B

C

Figure 34 a-c. Radiographs of the right hip of a 43-year-old female with a subcapital femoral neck fracture (Garden II) preoperatively (a), after fixation with three SR-PLLA lag screws (b) and two years postoperatively. The borders of the screw channels are still visible. The fracture has united (c).

A



B

C

Figure 35 a-c. Radiographs of the right hip of a 75-year-old female with a subcapital femoral neck fracture (Garden II) before operation (a), after fixation with three metallic screws (b), and 19 months postoperatively. The fracture has united (c).

A

B

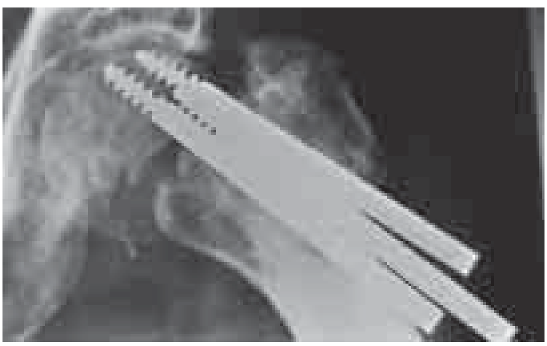
C

Figure 36 a-c. Radiographs of the left hip of a 65-year-old female with a displaced subcapital femoral neck fracture preoperatively (a), after reduction and bioabsorbable fixation (b) and one month postoperatively with redisplacement of the fracture (c).

Figure 37 a-c. Radiographs of the left hip of a 65-years-old male with a displaced subcapital femoral neck fracture preoperatively (a), after reduction and metallic screw fixation (b) and 10 months postoperatively with redisplacement and non-union of the fracture (c).

A

B



C

4.5. ABSORBABLE FIXATION OF FEMORAL HEAD FRACTURES. A PROSPECTIVE STUDY OF SIX CASES (V).

The results of this study were evaluated by clinical and radiological methods. The criteria recommended by Thompson and Epstein (1951) are still in use (Table 5 a – b). The clinical and radiological results were rated independently. If rating by the two methods differed, the case was assigned to the lower of the two grades (Table 6).

Table 5 a. Clinical criteria for evaluating results.

Excellent: All of the Following	*Good:*
1. No pain. 2. Full range of hip motion. 3. No limp. X No roentgenographic evidence of progressive changes.	1. No pain. 2. Free motion (75 per cent of normal hip). 3. No more than a slight limp. X Minimal roentgenographic changes.
Fair: Any One or More of the Following	*Poor:* Any One or More of the Following
1. Pain, but not disabling. 2. Limited motion of hip. No adduction deformity. 3. Moderate limp. X Moderately severe roentgenographic changes.	1. Disabling pain. 2. Marked limitation of motion or adduction deformity. 3. Redislocation. X Progressive roentgenographic changes.

(X = Roentgenographic criteria summarized)

Table 5 b. Radiological criteria for evaluating results.

Excellent (Normal)*:* All of the Following	*Good* (Minimal changes)*:*
1. Normal relationship between head and acetabulum. 2. Normal articular cartilaginous space. 3. Normal density, head of femur. 4. No spur formation. 5. No calcification in capsule.	1. Normal relationship between head and acetabulum. 2. Minimal narrowing of cartilaginous space. 3. Minimal de-ossificaton. 4. Minimal spur formation. 5. Minimal capsular calcification.
Fair (Moderate changes)*:*	*Poor* (Severe changes)*:*
1. Normal relationship between head and acetabulum. Any one or more of the following: 2. Moderate narrowing of cartilaginous space. 3. Mottling of head, areas of sclerosis, and decreased density. 4. Moderate spur formation. 2. Moderate to severe capsular calcification. 6. Depression of the subchondral cortex of the femoral head.	1. Almost complete obliteration of cartilaginous space. 2. Relative increase in density of the femoral head. 3. Subchondral cyst formation. 1. Formation of sequestrae. 5. Gross deformity of the femoral head. 6. Severe spur formation. 7. Acetabular sclerosis.

Table 6. Patients, methods and results.

Patient	1	2	3	4	5	6
Age (years)	20	61	21	29	21	46
Sex	male	male	male	male	female	male
Weight (kg)	65	126	62	110	55	110
Type (Pipkin)	2	2	2	2	1	4
Size of fragments of the femoral head	1/2	3/4	1/3	1/2	1/5	1/3
Reduction of dislocation of the hip	open	open	closed	closed	closed	open
Operation time elapsed from the injury (days)	24	1	3	0	0	0
Implants	SR-PLLA -screws	SR-PLLA -screws	SR-PLLA- and SR PGA rods	SR-PLLA- rods	SR-PLLA- rods	SR-PLLA- rods
Full weight bearing (weeks)	14	8	6	10	12	ín bed rest
Complications	myositis ossificans	aseptic necrosis	-	-	-	Died from cerebral injury
Follow-up (months)	77	12	55	48	36	6 weeks
Radiological result	fair	poor	excellent	excellent	excellent	-
Clinical results	good	fair	excellent	excellent	excellent	-

Case 1 with a delay of 24 days for fracture fixation had a follow up of six years and five months. The fracture was fixed with two SR-PLLA screws. At follow-up he had no pain, there was a slight limitation in flexion and he had a slight limp due to peroneal palsy caused by the accident. The radiographs showed ectopic bone formation in soft tissues anteriorly and posteriorly of the hip joint. The patient was reoperated on twice, 15 months and 27 months after the accident, to excise the calcifications which limited movement of the joint. The total sick leave was 17 months. He returned to his previous job.

In case 2 with a big main fragment consisting of three fourths of the femoral head, fixation was carried out retrograde through the femoral neck with one SR-PLLA lag-screw 6.3 mm in diameter and one fully threaded SR-PLLA screw 4.5 mm in diameter. Healing was uneventful in the beginning: radiographs showed bony union at eight weeks and there were no problems in mobilization. Six months after the operation he complained of increasing pain and MRI showed signs of avascular necrosis. Total hip replacement was performed one year after the injury. In the four other cases, fixation was performed by absorbable rods 2 mm in diameter. In case 3 the fracture was fixed with one SR-PLLA rod and one SR-PGA rod. The follow-up time was 4 years 7 months. The patient has full range of movement of the hip joint, no pain and no signs of osteoarthrosis or avascular necrosis. He had 14 weeks sick-leave and six months after the injury he returned to his former sport activities.

Case 4 had three main fragments, each of them fixed with two SR-PLLA rods. There was no pain and he had full range of movement of the hip. A minimal calcification can be detected radiologically. The follow-up was four years. He needed a sick-leave of 12 months because of his hard work. He has returned to his former sport activities.

The fracture in case 5 was fixed with four SR-PLLA rods. The female patient required sick leave of one year and five months. She has returned to sport activities and in the last follow-up had an excellent clinical and radiological result.

Case 6 was a patient with a severe cerebral injury, which caused his death six weeks post-operatively. The fracture was fixed by using three SR-PLLA rods. The radiographs six weeks postoperatively showed a good primary result. (Fig. 38-39).

Figure 38 a-b. Radiographs of the right hip of a 22-year-old male with a femoral head fracture fixed with SR-PLLA and SR-PGA rods 2 mm in diameter 6 months postoperatively (a) and 4,5 years after the operation (b). At 6 months both of the rod channels are visible. At 4,5 years the PGE channel has been filled with bone but the PLLA channel with sclerotic borders can still be seen.

Figure 39. A magnetic resonance imaging scan of the right femur of 24-year-old male 3,5 years after bioabsorbable rod fixation of a femoral head fracture. The channel of a 2 mm PLLA rod is still visible.

5. DISCUSSION

Bioabsorbable implants have been used successfully in the fixation of cancellous bone fractures and arthrodeses (Rokkanen et al. 1985, Partio et al. 1992a–d). Encouraged by these clinical results and those of experimental studies on the development of ultra-high strength bioabsorbable self-reinforced implants, subcapital femoral osteotomies in sheep were fixed with two SR-PLLA or two metallic screws to find out if bioabsorbable fixation was a suitable method for fixation when the area is exposed to high mechanical stresses. This experimental osteotomy could also act as a model for a subcapital femoral neck fracture in a human. Promising results were already achieved in a previous experimental study in which this type of osteotomy was fixed with three SR-PLLA pins (Vasenius et al. 1993). However, it was concluded that due to slight instability in a direction parallel to the axis, semithreaded implants are needed.

In the present study two screws were used for each fixation because of the narrow shape of the femoral neck in sheep; an intertrochanteric fossa does not leave enough space for three screws. The metallic screws were bigger in the outer diameter of the threaded part than the SR-PLLA screws, but the diameter of the unthreaded part was closer in size to the bioabsorbable screws than the smaller available metallic screws would be. The metallic screws were removed before testing, whereas degradation of the SR-PLLA screws made their removal impossible, and they were left in place. Moreover, the new bone formation around the threads seen macroscopically in cross sections kept the bioabsorbable screws in place. The effect of the remaining screws on the load carrying capacity of the osteotomy planes was difficult to estimate because, in addition to simple shear load, tensile, bending and compression forces also burden the implants. In our investigations, the initial bending strength of a SR-PLLA screw (257.9 MPa) decreased to 68.6% in 12 weeks. The fact that none of the bones from either group failed at the level of osteotomy showed that the union of the osteotomy was strong. The results of this study show that SR-PLLA screws are as safe as metallic screws in the fixation of subcapital femoral osteotomies in sheep.

Although PLLA implants have gained in popularity during the last two decades as an alternative fixation method to metallic devices, the previous investigations have not given us exact data on the degradation process and on the bone tissue response in the long term. The degradation time is long, and even presumably depending on many factors such as the geometry of the implant and the site of implantation, as already pointed out (Vasenius et al. 1990, Suuronen et al. 1998). Matsusue et al. (1995) investigated the degradation of PLLA rods 3.2 mm in diameter and 50 mm in length, manufactured by using the drawing technique, after intramedullary and subcutaneous implantation in their long-term study of rabbits. The intramedullary implanted rods had been absorbed almost completely at 62 months and completely by 69 months when implanted subcutaneously. After an intramedullary implantation, the PLLA rods had been replaced by bone marrow cells. In another experimental long-term study, SR-PLLA multiplayer plates were used to fix mandibular osteotomies in sheep. The plates were fixed with titanium screws. All the osteotomies healed. After five years, the material was almost completely resorbed, but small particles of polymer could still be detected at the implantation site. Although the material had degraded, the plates retained their macroscopic form until the end of the five-year follow-up (Suuronen et al.1998). In the present study about long-term degradation the screws were implanted in the bone. In histological examination at five years and seven years and four months all the PLLA material had disappeared. The cutting of the thin sections was extremely difficult which was probably due to the delay between killing the sheep and fixing the bones in ethanol immersion, due to the fact that it took two to three days before the CT and MRI examination of the specimen as whole legs could be carried out. For that reason, we also used 80 μm sections for microscopy and stained them with the Masson-Goldner trichrome method like normal sections, too. Also the thick sections were examined by light microscopy and polarized light to be sure, that the PLLA material was not only cut away of thin sections, but was also really biodegraded. At the time when we started the study no long-term studies of SR-PLLA were available and therefore 5 years was expected to be an appropriate follow-up time.

The SR-PLLA screws were replaced mostly by bone tissue with some connective tissue. The new bone was however more tight and in CT examination the density values were higher than these of the surrounding bone, corresponding to the values of compact cortical bone. In previous studies, PGA, which has a shorter degradation time, has been found to be replaced by cancellous bone. It is to be assumed that if we could accelerate the degradation of lactide, also the bone remodelling would take place earlier and could then lead to more cancellous bone like-tissue replacement.

The strength values of the SR-PLLA implants decreased gradually and at 12 weeks the bending strength was one-half of the initial values in a previous study (Majola et al.1992). In this study (III) the bending strength of the SR-PLLA screws 4.5 mm in diameter was not decreased and that of the 6.3 mm screws only 10.9 %. Respectively, the values at 18 weeks were 16.7% and 72.7 %. The explanation of the rapid strength retention of 6.3 mm SR-PLLA screws between 12 and 18 weeks may be the mechanical wear on long screws during implantation, which was also visible after preparation. Also the slight increase in bending strength from 27.3% up to 30.8% after six more weeks hints at some disturbance. The reported results for PLLA with different morphological structures show that a considerable loss of molecular weight can occur before there are significant changes in mechanical properties (Bos 1989), although a strong correlation between strength loss and molecular weight loss has also been shown in SR-PLLA rods in vitro (Pohjonen and Törmälä 1994). In our study, the loss of molecular weight tends to be more rapid and linear than strength retention in the first 12 to 18 weeks. The shear strength keeps its percentile values better than the bending strength up to 24 weeks (42% in 6.3 mm and 76% in 4.5 mm SR-PLLA lag-screws at 24 weeks).

It is possible to change mechanical properties during manufacturing even where the same material is concerned. In the drawing process, the bending and shear strengths increase almost linearly as a function of the draw ratio. The torque strength, which is especially important in clinical work, starts to diminish after a certain draw ratio when the structure becomes fully fibrillated.

In a study Pohjonen et al. (1997) compared SR-PLLA screws manufactured by the compression moulding technique and by an in-house developed machine cutting technique. The PLLA billets used for the manufacture of new machine cut SR-PLLA screws were partially fibrillated after uniaxial extension by 400%, while the material of old type SR-PLLA screws was completely fibrillated after uniaxial extension by 700%. New machined SR-PLLA screws were significantly stronger in bending and torque strength tests but significantly weaker in shear strength tests. The optimising of the strength properties of the material to the suitable devices still needs further investigation in the future. Therefore, in the development of implants, the continuous interaction of engineering and medical sciences is important.

To the knowledge of the author, bioabsorbable screw fixation has not been performed earlier in the treatment of subcapital femoral neck fractures. In experimental studies SR-PLLA screws have proved to be comparable to metallic screw fixation also in the weight bearing area (Paper I). The first patient of the present study was operated on in January 1991, using SR-PLLA lag screws. After two failures of the fixation in Garden IV fractures it was decided to limit further patients in this first study performed by a new method to those with Garden I, II and III fractures. We compared the results achieved with bioabsorbable fixation to those with metallic screw fixation in the patients treated during the same time period at our department. The pre-operative dislocations of the fractures were analysed and divided into various stages of Garden's classification by the same orthopaedic surgeons in both groups to avoid variation in the criteria (Frandsen et al. 1988, Oakes et al.2003). From the radiological and clinical aspects, the results of bioabsorbable screw fixation have been comparable with those of metallic fixation. The main advantage after bioabsorbable screw fixation has been the avoidance of the removal of metallic screws due to migration or screw backing with subcutaneous irritation or, if there is a need, for later surgery. The method also makes the use of MRI possible. These results justify recommending the use of the bioabsorbable SR-PLLA lag-screw fixation in Garden stage I and II fractures and selectively in young patients with Garden stage III fractures.

The benefits of biodegradable fixation are emphasized in intra-articular fractures with small or multiple fragments because biodegradable rods offer the possibility to fix small osteochondral fragments even directly through the cartilage surface. In a recent study of 30 femoral head fractures, the treatment was in 14 cases excision of the fragment(s) because of the size of the fragments, which were too small for screw fixation and 3 patients were treated with a hemiarthroplasty. One out of 14 patients treated by excision of the fragment(s) had an excellent clinical result according to Epstein's criteria, whereas in the group treated by internal fixation six out of 13 patients reached excellent clinical results (Yoon et al. 2001).

A posterior approach has been preferred in posterior fracture dislocations because of blood supply conditions (Epstein et al. 1985), but a later study has contradicted these conditions, showing no risk of increased avascular necrosis of the femoral head following an anterior approach (Swiontkowski et al. 1992). CT scan or MRI may help to decide whether a posterior or anterior approach should be chosen (Kuhn and Frymayer 1987, VerMeiren and Haye 1991, Guiral et al. 1992, Simmert et al. 1993). In this study, one hip was operated on through a lateral approach, CT examination showing the fragments on the anterior aspect of the femoral head. The reattachment of the ligamentum teres has been discussed in connection with operative treatment of this trauma to decrease the rate of avascular necrosis. Although the main blood supply is via the retinacular vessels, which arise from an extracapsular arterial ring formed by the medial and lateral circumflex arteries whereby the posterior-superior retinacular arteries from the lateral epiphyseal vessels supply most of the femoral head. The arteries of the ligamentum teres supply a small area of bone around the fovea centralis. The revascularisation of the femoral head after a fracture also occurs from the medullary cavity.

The active development of bioabsorbable bone fixation devices has raised increasing interest on bioabsorbable implants in other surgical and medical fields also. In orthopaedics, PLLA cages for lumbar interbody fusions (van Dijk et al. 2002, Toth et al. 2002) and implants for lumbar intervertebral disc replacement (Palmgren et al. 2003) have been developed. Bioabsorbable SR-PLLA stents have been investigated in urology (Talja et al. 1997, Isotalo et al. 2002) and in vascular surgery (PLA96) (Hietala et al. 2001 and 2003). Bioabsorbable material has also been used for scaffolding the peripheral nerve regeneration (Ngo et al. 2003) or bone substitution material (Ignatius et al. 2001). Bioabsorbable composite materials can also act as carriers for drugs, such as antibiotics or chemotherapy for tumours (Brem and Habikian 2001, Rutledge et al. 2003), hormones (Ma et al. 2000) or growth factors (Tielinen et al. 1998, 1999).

The development of bioabsorbable materials started with an idea of replacing metallic implants because of some superior properties. The most valuable studies have been randomised trials where novel implants have been compared with metallic fixation devices – the gold standard. And the operation techniques have also frequently imitated those developed for traditional metal implants. On the other hand, despite of the availability and acceptance of bioabsorbable devices, many surgeons are unfamiliar with the new technology. However, the intense research and development of biodegradable materials in past decades has led us to situation where new implants in many indications cannot be replaced by traditional hardware.

6. CONCLUSIONS

1. Mechanical properties of SR-PLLA screws in the fixation of experimental subcapital osteotomies in sheep have proved to be comparable to metallic screws with a high rate of union.

2. Histologically, the connective tissue started to penetrate into the implant at one year and at two years SR-PLLA lag-screws were fragmented and at three years small islands of the implants were seen. At three years the implant was split into small fragments, surrounded by connective tissue, while a mild to moderate foreign-body reaction was observed. At five years and at seven years and four months the implant material was totally biodegraded.

3. The initial strength values of SR-PLLA lag-screws were high. At twelve weeks the strength values were not marked decreased and at 36 weeks the screws had only 15 % of their initial strength. The gradual strength retention during the biodegradation process makes the use of them safe even in demanding fixations.

1. This preliminary report showed that SR-PLLA lag-screws can be used safely to fix subcapital femoral neck fractures in Garden I and II fractures and selectively in younger patients with Garden III fractures. There were no statistical differences in the outcome of the patients treated with metallic or absorbable implants.

2. SR-PLLA and SR-PGA implants offered marked advantages in the fixation of femoral head fractures associated with a traumatic dislocation of the hip joint. These intra-articular fractures with usually small fragments can be fixed even through the cartilage surface and there is no need for an implant removal procedure later on. The biocompatibility of the implant material used appeared high.

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